Increasing engagement during robot-aided motor rehabilitation using augmented feedback exercises

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Publication Date:
2013

Permanent Link:
https://doi.org/10.3929/ethz-a-010002943

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Increasing Engagement during Robot-Aided Motor Rehabilitation using Augmented Feedback Exercises

A dissertation submitted to
ETH ZURICH

for the degree of
Doctor of Sciences

presented by
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February 23, 1983

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2013
Acknowledgements

During my joint PhD at ETH Zurich and Hocoma AG, I was lucky to have the support of many people. The combination of scientific research and industrial product development allowed me to directly apply newly acquired knowledge on products that are being used by patients in clinics around the world. Therefore, I first want to thank those people that made this possible. This includes Robert Riener from ETH Zurich as well as Lars Lünenburger and Gery Colombo from Hocoma AG.

Thanks Robert for giving me this possibility, the full support during the last couple of years and for all the great advice, feedback and great times during and besides my work at ETH. Thank you Lars for your guidance during my thesis, all your advice, support, feedback and the great times we had during work. Thank you Gery for giving me your confidence, the full support and the possibility to pursue my PhD during my work at Hocoma AG. I further want to thank Roger Gassert, my co-advisor from ETH Zurich, for the great and fruitful feedback, comments & discussions.

Special thanks also go to Marc Bolliger, Peter Wolf & Alexander König. Thanks a lot for all your time, for all your advice, feedback and your input during my thesis.

I want to thank all my colleagues at Hocoma AG as well as the Sensory-Motor Systems Lab and RELab of ETH Zurich, for the laughters and the great collaborations we had during my thesis.

I want to thank Carmen Krewer and Mario Jacky for their great effort in helping me during my studies as well as their valuable inputs and feedback.

A big thanks also to all of my friends. Thank you for the great evenings, the inspiring, interdisciplinary discussions, the laughters and all the times we shared.
Last but not least, I deeply want to thank my mother, Therese, the most warm-hearted and caring person I know, my father, Christian, who I admire for his passion and wealth of knowledge, and my sister, Sonja, who inspires me with her enthusiasm and creativity. Thanks for your infinite support in pursuing my dreams, for all your love, encouragements and being there for me during the past years. I couldn’t have done this without you!
Abstract

Motor functions are essential for socio-ecological participation. Impairments, resulting from spinal cord injuries, stroke or other cerebrovascular diseases, influence a persons’ independence and, thus, require restoration, through rehabilitation.

Due to the demanding implications of conventional therapy approaches, robotic interventions have become increasingly accepted. However, despite all their advantages, robotic assistance can reduce patient engagement during training. The primary cause for this effect can be attributed to the passive guidance that the robot provides, preserving desired movement kinematics and reducing actual movement errors. The absence of errors eventually leads to a reduction of effort, which can influence the overall effectiveness of the therapy.

Patient engagement can be defined as a construct that is driven by motivation and executed through active, effortful participation during therapy. It is a multidimensional construct that contains cognitive, physical and emotional components. Cognitive engagement may be influenced by a person’s cognitive abilities in relation to task specific characteristics, like for example, the level of difficulty. Physical engagement may be influenced by physiological impairments. The emotional component of engagement mostly depends on the psychological state of the patient, being influenced by, for example, preferences, expectations or satisfaction.

In order to increase patient engagement, modern therapy approaches deploy “game-like” augmented feedback exercises, making use of virtual reality technology, to create multi-sensory, virtual therapy scenarios. Most exercises, however, are not properly verified regarding their effectiveness. In addition, although some studies did propose design characteristics that augmented feedback exercises should inherit to optimally
enhance engagement, these propositions are mainly based on the evaluation of questionnaires and, thus, their actual effectiveness to increase engagement during therapy has not been verified.

Hence, the overall aim of this thesis was to assess the effect that augmented feedback exercises have on motivation and investigate what characteristics they need to include, to increase active participation, that is, patient engagement, during robot-aided motor rehabilitation.

Since there are no explicit guidelines of how augmented feedback exercises have to be designed, I postulated that they should be able to adapt their level of difficulty to the capabilities of the patient. Further, augmented feedback exercises should provide explicit task goals, making therapy benefits apparent, provide frequent performance feedback and allow adjustable visual stimuli. Finally, they should show functionally meaningful reactions to the motor performance of the patient and allow the possibility to accomplish competitive training. The effects of these design characteristics on patient engagement, were then assessed in both upper- and lower-extremity robotic devices.

In general, results showed an increase in engagement of patients during therapy using augmented feedback exercises. This was not only reported by patients, but also perceived by their therapists. Patients were motivated to train using augmented feedback exercises and wanted to use them again in subsequent therapy sessions.

Results further showed that it is difficult to manually adjust the task difficulty of an augmented feedback exercise to the capabilities of the patient. Although patients showed an increase of effort with increasing task difficulty, they perceived their performance as getting worse, which eventually led to increased levels of frustration. In order to prevent the possibility of disengagement, augmented feedback exercises should, thus, be able to autonomously adapt their level of difficulty to the capabilities of patients.

I was able to successfully validate such a mechanism, that, thanks to its simplicity, can be easily incorporated into a large number of existing and novel upper-extremity, two-dimensional, pointing exercises.

Finally, I was able to show that augmented feedback exercises have to provide functionally meaningful reactions to the motor performance of the patients, in order to
increase active participation. Providing explicit task goals, frequent performance feedback or competitive situations did not have any significant effects on engagement. The effects of different visual stimuli on patient engagement were not assessed in the current thesis and, thus, should be the subject to further investigations.

In conclusion, deliberating on the multidimensional construct of patient engagement, I argue that cognitive engagement can be increased by incorporating mechanisms into augmented feedback exercises that autonomously adapt their level of difficulty to the capabilities of the patient. In addition, the amount of physical engagement patients exhibit during therapy can be increased by augmented feedback exercises that provide functionally meaningful reactions to the motor performance. While the emotional component of engagement was not directly assessed within the current studies, patients did show differing preferences and expectations towards the developed augmented feedback exercises. These were not only related to the visual setup of the virtual environments, but also to the overall tasks themselves. Hence, while results did not show significant effects regarding task goals, performance feedback or competitive situations, these could potentially have an influence on an individual patient’s level of emotional engagement.

At last, I encourage further investigations on the effects that different design characteristics have on engagement during robot-aided motor rehabilitation. Different aspects will be highlighted and discussed in the outlook section of this thesis.
Zusammenfassung

Um aktiv am Geschehen unseres sozialen Umfeldes teilnehmen zu können, spielen motorische Funktionen eine wichtige Rolle. Diese können jedoch auf Grund von Rückenmarksverletzungen, Schlaganfällen oder anderen cerebro-vaskulären Schäden stark beeinträchtigt werden. Die Neurologische Rehabilitation kann in solchen Fällen zu signifikanten Verbesserungen der Motorischen Fähigkeiten führen und dadurch die Eigenständigkeit und dieIntegration von betroffenen Personen in die Gesellschaft erleichtern.


Patienten-Engagement ist ein multidimensionales Konstrukt welches durch die Motivation angetrieben wird, und zur aktiven Teilnahme während der Therapie führt. Es beinhaltet sowohl kognitive, physische als auch emotionale Bestandteile. Kognitives Engagement kann durch die vorhandenen kognitiven Fähigkeiten eines Patienten in Bezug auf die Eigenschaften einer Aufgabe, wie zum Beispiel deren Schwierigkeitsgrades, beeinflusst werden. Allfällige physiologische Störungen wirken sich vor allem
auf den physischen Bestandteil von Engagement aus. Der psychische Zustand des Patienten sowie dessen, Erwartungen, Präferenzen etc. steuern den emotionalen Bestandteil.


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Introduction

According to a survey by the “Christopher Reeve Foundation” amongst 5.6 million people living with paralysis, 29% were caused by stroke, 23% were caused by spinal cord injuries (SCI) and 17% by multiple sclerosis. Out of all people that reported being paralyzed, 36% reported having “a lot of difficulty” in moving their upper or lower extremities, 29% reported “some difficulty”, 17% “a little difficulty” while 13% reported to be “completely unable” to move (Christopher & Dana Reeve Foundation, 2011).

Being able to move, i.e., motor functions, however, are essential for socio-ecological participation. Impairments hence influence a persons’ independence and, thus, require maximum possible restoration, through neurological motor rehabilitation, in order to improve an affected person’s quality of life (Rösser et al., 2008).

1.1 Neurological Motor Rehabilitation

Within the field of neurological motor rehabilitation, the general assumption is, that the recovery of an affected movement has to be stimulated by an intensive training that aims at the function to be regained (Willingham, 1998, Dobkin, 2003, Lünenburger et al., 2004). Such training is believed to cause undamaged brain regions to take over the generation of muscle commands to the same muscles as were used before the injury, termed as neural plasticity (Nudo, 2006). Apart from spontaneous recovery, a biological process that improves performance across a range of tasks in a short time
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after injury, the degree of performance improvement, is believed to be dependent on the amount of practice during therapy (Kwakkel et al., 1997). Thus, intensive practice and frequent repetitions of the motor patterns that have to be regained are both integral parts of motor rehabilitation.

1.1.1 Robot-Aided Motor Rehabilitation

During conventional gait therapy approaches, therapists manually assist affected limb movements in a physiological way (Colombo et al., 2000). Due to possibly ergonomically unfavorable positions of the therapists and the way they have to assist the patient’s movements, conventional therapy approaches can, however, be quite demanding and tiring for both patient and therapist, reducing the overall duration and the intensity of therapy (Hidler and Wall, 2005, Colombo et al., 2000, Riener et al., 2011). Hence, in recent years, the use of robotic interventions, i.e., external orthotic appliances, that prevent or assist the movement of limbs, have become increasingly accepted in the field of rehabilitation (Colombo et al., 2000, Sanchez et al., 2004, Prange et al., 2006, Kwakkel et al., 2008, Housman et al., 2009). Devices like, e.g., the MIT-Manus (Krebs et al., 1999), the T-WREX (commercialized as Armeo® Spring, Hocoma AG, Switzerland) (Housman et al., 2007, 2009), the Bi-Manu-Track (Hesse et al., 2005) or the ARMin (commercialized as Armeo® Power, Hocoma AG, Switzerland) (Nef et al., 2009) are used to train upper-extremity movements. Examples of robotic interventions for lower-extremity rehabilitation, include the Lokomat (Colombo et al., 2000, Riener et al., 2010), the WalkTrainer (Stauffer et al., 2009), or the LOPES (Veneman et al., 2007). These upper- and lower-extremity devices provide constant and secure guidance of affected limbs (Fasoli et al., 2004, Hidler and Wall, 2005, Lünenburger et al., 2004), allow longer and more frequent training sessions of higher intensity (Hidler et al., 2005, Riener et al., 2005, Brewer et al., 2013) and reduce the labor-intensive workload of therapists during rehabilitation on a daily basis (Lum et al., 2002, Colombo et al., 2001, Wirz et al., 2005).

Despite all the advantages of robotic interventions, robotic assistance can reduce a patients’ physical effort to actively participate during training. This has been shown for both upper- and lower-extremity therapy (Israel et al., 2006, Wolbrecht et al., 2008).
1.1 Neurological Motor Rehabilitation

The primary cause for this effect can be attributed to the passive guidance that the robot provides during training (Israel et al., 2006). Continuous guidance preserves desired movement kinematics, reducing actual movement errors. Since the human motor system attempts to decrease muscle activation during repetitive movements when errors are small, this effect leads to slacking (Reinkensmeyer et al., 2009, Marchal-Crespo and Reinkensmeyer, 2009), i.e., it minimizes physical effort and, thus, can negatively affect motor learning and the overall effectiveness of the training (Kaelin-Lang et al., 2005, Secoli et al., 2011). Since active participation, however, constitutes an important factor for successful rehabilitation (Langhorne et al., 1996, Lotze et al., 2003), robotic therapy still relies on physiotherapists to keep patients motivated and engaged.

1.1.2 Motivation & Engagement

Motivation can be defined as the “forces acting on or within a person to initiate a behavior” (Phillips et al., 2004) or even simpler “the direction and intensity of ones effort” (Sage, 1971). Psychology, neuroscience or rehabilitation are only some of the many research fields that actively address human motivation in order to improve, e.g., working performance, understand social behavior or try to improve therapy outcomes (Maslow, 1943, Vroom, 1964). Motivation can be subdivided into intrinsic and extrinsic motivation. Intrinsic motivation can be described as “the doing of an activity for its inherent satisfactions rather than for some separable consequence” (Ryan and Deci, 2000), i.e., intrinsic motivation occurs if people do something that they enjoy (e.g., through curiosity, exploration or play). Extrinsic motivation, in contrary, can be seen as a construct that “pertains whenever an activity is done in order to attain some separable outcome” (Ryan and Deci, 2000), i.e., it can be influenced by some sort of external reward (e.g., a prize or money). While people at first might only participate in an activity because of a subsequent reward, i.e., extrinsic motivation, they over time may start to experience the activity’s intrinsically interesting properties, which may eventually cause an orientation shift from extrinsic to intrinsic motivation. This “internalization” can be facilitated by activities that support competence (e.g., optimal challenges or effectance-promoting feedback) (Ryan and Deci, 2000). In motor rehabilitation, patients already have a personal, extrinsic motivation to regain their movement capabili-
1. Introduction

This motivation could, however, further be increased by turning the mostly boring, repetitive trainings into enjoyable and entertaining therapy sessions.

Building upon motivation, engagement can be defined as “the act of beginning and carrying on of an activity with a sense of emotional involvement or commitment and the deliberate application of effort” (Lequerica and Kortte, 2010). Hence, engagement can be seen as a multidimensional construct that contains cognitive, behavioral, i.e., physical and emotional components (Kahn, 1990, Lequerica and Kortte, 2010). Cognitive engagement may be influenced by a person’s cognitive abilities (e.g., perception, attention, memory) in relation to task specific characteristics (e.g., the level of difficulty). Physical engagement is reflected through participation and may be influenced by physiological impairments. Finally, emotional engagement can be influenced through a person’s current mood, preferences, expectations or satisfaction (Macey and Schneider, 2008).

Motivation and engagement, thus, play key roles when it comes to motor learning. For motor rehabilitation we can define patient engagement as a construct that is driven by a patients’ motivation, i.e., effort to regain movement capabilities and executed through active, effortful participation during therapy (Lequerica and Kortte, 2010) resulting in increased physical, i.e., muscle activation. In addition to physical engagement, cognitive engagement during motor rehabilitation can be increased by relating the difficulty of a motor learning task to the cognitive abilities of the patient. Since the emotional component of engagement mostly depends on the psychological state of the patient, it is the most difficult to influence. Active participation during therapy, therefore, can most effectively be influenced through the cognitive and physical aspects of engagement. One method that has been shown to increase motivation and, thus, promote engagement during motor learning is movement feedback (Timmermans et al., 2009).

1.1.3 Movement Feedback

Feedback allows patients to evaluate their movement success as well as detect potential movement errors (Van Vliet and Wulf, 2006, Subramanian et al., 2010). Concepts that are mostly being deployed are, e.g., verbal feedback and mirrors placed in front of the patients, giving visual and/or acoustic feedback (Dobkin et al., 2003). Other
approaches include biofeedback, i.e., processes that monitor a patient's performance using psycho-physiological measurements (Horowitz, 2006).

While feedback about the quality of movement has been shown to be beneficial for effective motor learning, one, however, should always keep the patients’ neurological deficits in mind, when choosing a suitable type of feedback, in order to prevent overloading patients' perceptions due to possible impaired perceptual and cognitive abilities (Willingham, 1998, Boyd and Weinstein, 2003, Huang et al., 2006, Van Vliet and Wulf, 2006).

With the advent of modern media technologies, the use of another sort of feedback, augmented feedback (AF) has nowadays become increasingly popular (Schmidt et al., 1989, Magill, 2004, Holden, 2005, Chiviacowsky and Wulf, 2007). AF can be defined as “the provision of supplementary sensory information (visual, auditory, or haptic) brought by technological means, which would not normally be present in the usual environment” (Didier and Bigand, 2010). Hence, movement sensors can, for example, be used to record small bodily movements that might not be visible otherwise and augment them through auditory (e.g. guidance sounds) or visual (e.g. computer simulations) feedback (see Table 1.1). Adding haptic feedback, further allows for manual interactions with the AF exercise (Bresciani et al., 2008). The next section will thus focus on the implications of using AF to increase motivation and engagement during therapy and highlight important drawbacks of today’s approaches.

1.2 Augmented Feedback Exercises & Virtual Reality

AF exercises incorporate digital displays (e.g., computer screens), making use of the advantages of virtual reality (VR) technology (Linguist, 2002, Magill, 2004, Liebermann et al., 2006) to create multi-sensory, computer generated simulations of both fictional and genuine environments, i.e., virtual environments (VE). Subjects interact with VEs through different kinds of human-machine interfaces (e.g., visual display devices, sensors, haptic/force feedback devices, audio devices, input devices) (Sveistrup, 2004, Holden, 2005). Newer approaches even use physiological signals from the human
1. Introduction

body (e.g., heart rate, blood pressure, body temperature, skin conductance) as interaction modalities for VEs (Andreassi, 2006, König et al., 2011b). In lower-extremity rehabilitation using treadmills, AF exercises are generally presented on monitors in front of the patient. Newer approaches, however, have also started to project the AF exercise directly onto the instrumented treadmill, allowing to train foot positioning relative to environmental properties (van Ooijen et al., 2013).

Different therapeutic contexts have all already made use of AF technology (Riva et al., 2006). VEs do not only seem to have a lot of potential for memory training (Guo et al., 2004), but they are also used to treat multitasking deficits (Rand et al., 2009), dementia (Flynn et al., 2003), obesity (Riva et al., 2001, Lu et al., 2013) or assessing navigational skills in persons with alzheimer’s disease (Cushman et al., 2008). Other contexts were VEs are deployed include pain reduction in burn patients (Hoffman et al., 2000), phobia treatment (De Carvalho et al., 2010), treatment of post-traumatic stress disorders (Rothbaum et al., 2010, Rizzo et al., 2009, McLay et al., 2012), or rehabilitation of attention deficit hyperactivity disorders (Rizzo et al., 2006). More recent approaches use VEs to improve patients motivation during functional low back pain therapy, providing therapists with direct assessments of trunk movement quality (Valedo® Motion, Hocoma AG, Switzerland).

In motor rehabilitation, patients interact with VEs using their external orthotic appliances or measured biofeedback values. In addition to providing patients with concurrent (i.e., during the task) or terminal (i.e., after the task), precise, external movement performance feedback, AF exercises using VR in motor rehabilitation can be used to create different virtual rehabilitation scenarios with challenging, but safe, virtual therapy goals (Holden, 2005, Rizzo and Kim, 2005, Adamovich et al., 2009). Through game-like characteristics (see Table 1.1), AF can additionally increase the overall motivation and engagement during the mostly boring and repetitive trainings of long duration (Keshner, 2004, Subramanian et al., 2007, Perry et al., 2011).
1.2 Augmented Feedback Exercises & Virtual Reality

Table 1.1: Video Game Characteristics - Examples of game-like characteristics that augmented feedback exercises can make use of to increase the overall motivation and engagement during motor rehabilitation.

<table>
<thead>
<tr>
<th>Characteristic</th>
<th>Description</th>
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<tbody>
<tr>
<td>Characters</td>
<td>Create an external focus which has positive effects on performance (Chiviacowsky et al., 2010)</td>
</tr>
<tr>
<td>Reward System</td>
<td>Keep patients engaged (Lövquist and Dreifaldt, 2006)</td>
</tr>
<tr>
<td>Scores</td>
<td>Increase motivation (Burke et al., 2009)</td>
</tr>
<tr>
<td>Progress Bars</td>
<td>Indicate the progress towards a predefined movement goal (Burke et al., 2009)</td>
</tr>
<tr>
<td>Leader Boards</td>
<td>Provide patients with normative feedback, i.e., allow patients to compare their performance to that of others (Lewthwaite and Wulf, 2010)</td>
</tr>
<tr>
<td>Sounds</td>
<td>Guide a patient's movement (Huang et al., 2005)</td>
</tr>
<tr>
<td>Competition</td>
<td>Allow performance comparison (Vorderer et al., 2003)</td>
</tr>
<tr>
<td>Level of Difficulty</td>
<td>Match task difficulty to patient capabilities to increases engagement (Hill et al., 1985)</td>
</tr>
</tbody>
</table>

1.2.1 Augmented Feedback during Motor Rehabilitation

Different AF exercises using VR have already been developed for both commercial and research purposes. In upper-extremity motor rehabilitation, for example, AF exercises have shown to promote recovery in patients suffering from stroke (Piron et al., 2005, Levin et al., 2010) and lead to smoother arm movements (Huang et al., 2005). Other exercises again make use of neurobiological concepts like, e.g., the “mirror neuron” hypothesis, to improve current AF approaches (Cameirao et al., 2007, Siekierka et al., 2007, Eng et al., 2007b,a, Cameirao et al., 2009, 2010). In lower-extremity rehabilitation, AF exercises, amongst others, have been used detect and control gait freezing of patients with Parkinson’s disease (Baram et al., 2002, Bachlin et al., 2010, Espay et al., 2010), increase the physical engagement of children with cerebral palsy (Patritti et al., 2010, König et al., 2008a,b, Brütisch et al., 2010, Schuler et al., 2011), motivate subjects suffering from stroke (Zimmerli et al., 2009) and improve dynamic balance control (Jacobson et al., 2001, Betker et al., 2007). Designed exercises differ in the use of audio features, level of graphical detail (i.e., from simple to realistic), feedback presentation, level of immersion, and game mechanics.

1.2.2 Drawback of today’s approaches

Due to the increasing number of patients, the pressure of the healthcare system to reduce medical costs and limited therapeutic resources, clinicians may in the future want patients to practice their movements at home or with reduced supervision during
1. Introduction

their stay in the clinic. The need for AF exercises that are capable of keeping patients engaged and motivated will hence rise.

Despite the large amount of AF exercises that have been developed for motor rehabilitation, most of them are not properly analyzed regarding their effectiveness to increase engagement for motor learning during therapy (Edmans et al., 2006, Flores et al., 2008). Hence, most studies in research only assess the deployment effect of AF exercises, i.e., the effect that using or not using AF exercises has on engagement, or use AF exercises as an amendment to robotic appliances. Design characteristics that AF exercises should inherit, in order to optimally enhance engagement have not been addressed, i.e., there are no explicit guidelines of how such exercises have to be designed (Yim and Graham, 2007, Flores et al., 2008). In addition, although it is generally accepted that AF exercises effectively enhances motor learning, there are many controversial discussions about what kind of modalities (e.g., visual, auditory, or haptic) should be used and when feedback should be given for most effective learning (Sigrist et al., 2013).

Although some studies did propose characteristics that should encourage motivation, these are mainly based on the evaluation of questionnaires and, thus, their actual effectiveness to increase engagement during therapy has not been verified. According to Lövquist and Dreifaldt (2006), AF exercises should include a reward system to keep patients interested, allow the adjustment of the level of difficulty, provide real-time, task-specific feedback and build upon intuitive tasks. These were later extended by Pareto et al. (2008), adding the need for variable content.

While most AF exercises mainly make use of visual and auditory modalities, haptic feedback mostly gets neglected (Bresciani et al., 2008). Haptic feedback, however, especially through error augmentation where movement errors are temporarily magnified to encourage learning and compensation, have also been shown promising for motor recovery (Patton et al., 2006). Studies have further shown, that the congruency of real and virtual limb positions may affect the formation of visual consciousness (Salomon et al., 2013). Patients, thus, may benefit from accurately and congruently mapped augmented artificial bodies, i.e., avatars to relearn motor functions (Blanke, 2012). When using avatars, it, however, is not only the congruent mapping that affects motor learning but also the chosen viewport, i.e., the perspective of how patients
1.3 Characteristics of Video Games

Video games are known for their motivational and sometimes even addictive power to keep people engaged and immersed for a long period of time. Successful video games communicate outcomes of player actions in a perceivable way, not only having an immediate effect on the game, but also affect the playing experience at a later stage (Salen and Zimmerman, 2003, Burke et al., 2009). Using modifications of different video games, Malone (1980) identified three distinctive properties that video games should deploy to increase engagement. These are creating challenging environments where the outcomes are uncertain, stimulating fantasies as well as providing an optimal level of sensory and cognitive information complexity (Malone, 1980, 1981, 1982). While the underlying principles of game-play probably haven’t changed since Malone’s postulation, recent technological advancements, leading to realistic visual and auditory renderings, have resulted in higher expectations towards video game development.

Hence, Wood et al. conducted an online study about different video game characteristics, asking participants to rate their importance for enjoyment (Wood et al., 2004). Realistic sound effects were deemed to be highly important, while background music, speaking characters and narration were seen as less relevant or even distracting. Most participants preferred realistic, high quality graphics in realistic settings. Participants
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preferred medium duration games i.e. games that took days or weeks to complete, and the possibility of multi-player gaming. In addition, participants wanted to be able to choose between different levels of difficulty, sound and graphics settings. Regarding game dynamics, features like exploration, elements of surprise, fulfilling a quest or collecting and avoiding things, were all very popular. Finally, respondents rated points accumulation and finding bonuses as important, losing points as a necessity and being able to save and restart the game at regular intervals crucial.

Thus, Wood et al. were amongst the first to determine different preferences and expectations that users have towards video game playing. These were later revised by King et al. (2010), subdividing them into five categories:

(1) **Social features:** Elements that players can use to encourage each other, cooperate, compete or exchange strategies.

(2) **Manipulation and control features:** Elements that give the player control over the game, influencing a player’s sense or mastery.

(3) **Narrative and identity features:** Elements like the video game genre or game story.

(4) **Reward and punishment features:** Elements that reinforce playing and keep players connected to the game.

(5) **Presentation features:** Graphics and audio features.

However, one should also not forget the individual preferences amongst players. Some players enjoy video games because of the game story, graphics and sound. Hence, they neither have an interest in comparing their personal performance nor competing with other players, but rather play games for their personal amusement. In contrast, other players are more motivated to play because of the social experience, leader boards or because they prefer good game play over high-quality sounds and graphics. While some players loose track of how long they have been playing, others set strict personal time limits (Lazzaro, 2008, Westwood and Griffiths, 2010). This mirrors the huge amount of different game genres that exist, allowing players to choose those games that suit them the most.
Finally, studies were further able to show that certain video game characteristics can elicit emotional reaction in the player. Since psychological customization of video gaming is still in its infancy, not many studies exist up till now (Saari et al., 2004). Nevertheless, higher levels of difficulty, e.g., have been shown to elicit higher arousal (Ravaja et al., 2004, Tijs et al., 2009). While rewarding game events elicit positive emotional responses, negative events, during active participation, elicit increased positive affect (Ravaja et al., 2006). Similarly, different sound effects as well as music have been shown to trigger emotional responses and regulate the emotional state respectively (Bradley and Lang, 2000, Bernardi et al., 2006, Klimmt et al., 2009).

Despite all these findings, the characteristics described in this section were never examined in impaired subjects. Hence, while most of these concepts could also be valid during neurological motor rehabilitation, they should be addressed in more detail.

### 1.4 The Flow Theory

Another concept that has frequently been studied in association with media and video game entertainment, is the notion of flow.

The flow theory by Csikszentmihalyi (1975) describes different characteristics that make an experience enjoyable. Flow describes the state, when one does an activity for the “sheer sake of doing it” (Csikszentmihalyi, 1975, 1991, 1992, Nakamura and Csikszentmihalyi, 2005). It can be seen as a state of total engagement, so rewarding that people spend a great deal of energy, simply to be able to feel it (Csikszentmihalyi, 1975, 1992, Sweetser and Wyeth, 2005). Conditions for flow include, that an activity provides

(a) skill-dependent levels of difficulty,

(b) clear goals,

(c) and immediate progress feedback.

With these characteristics given, one can enter a subjective state of engagement with
1. Introduction

(a) an intense, i.e., focused concentration on the task at hand,
(b) a merging of action and awareness,
(c) a loss of the awareness of oneself,
(d) a sense of control over one’s activity,
(e) and a distortion of the temporal experience, i.e., an altered sense of time.

Hence, looking at these conditions, one can see analogies to the requirements for effective motor rehabilitation, namely, level of difficulty, goals and feedback. Since the flow theory describes essential characteristics that make an experience engaging, I claim that the application of flow to AF exercises for motor rehabilitation will not only increase the motivation of patients to actively participate, but also, influence their willingness to commit to longer and more frequent therapy sessions. The next sections will, thus, highlight how the flow theory has already been used by different research fields and how it could be applied to AF exercises.

1.4.1 Flow and Video Game Entertainment

Originating from studies about creativity, the flow theory has since been applied to many different research fields (Pearce and Howard, 2004, Reid, 2004). Likewise, game researchers have revisited the original flow components for video games (Jones, 1998, Sweetser and Wyeth, 2005, Cowley et al., 2008, Nacke, 2009). Sweetser and Wyeth (2005), for example, used the flow theory to develop the notion of GameFlow, by determining the elements of flow that are manifested in today’s video games. According to their model, games will be more absorbing, the more concentration is required, e.g., by creating tasks that are sufficiently challenging in comparison to a player’s skill level. Video games should further provide detailed game worlds that draw the player into the game, support skill development and provide a sense of control over the environment. They should react to the player, remembering all actions that were done during gameplay, provide explicit goals, and give direct feedback about the progress that has been made. Finally, video games should be developed in such a way, that
1.4 The Flow Theory

players feel immersed and create opportunities for social interaction, e.g., competition or cooperation. Although social interactions are not directly postulated by the flow theory, studies have shown, that including social components may further support pleasant game experiences (Vorderer et al., 2003, Gajadhar et al., 2008, Nacke, 2009).

Common to all “flow-to-game” adaptations, however, and manifested in most of today’s video games, is the possibility of adjusting the level of difficulty to the individual player. Hence, studies have shown, that individuals only engage in an activity, if the outcome matches the effort at which they perform (Hill et al., 1985, Kukla, 1972, Wright, 1982). In addition, individuals who believe that they are competent or successful, have been shown to remain engaged and motivated over a longer period of time (Lequerica and Kortte, 2010). Possible methods are having the player select a level of difficulty, performing game calibrations before play (Flatla et al., 2011), using dynamic difficulty adjustment during play (Hunicke and Chapman, 2004, Charles et al., 2005, Togelius et al., 2006) or by providing in-game events, where the player can choose to either increase or decrease the difficulty of the game (Chen, 2007).

To measure flow, conventional approaches include different types of questionnaires (Pearce and Howard, 2004, Weber et al., 2009) or the experience sampling method (ESM) (Gaggioli et al., 2003). During ESM, participants have to carry an electronic beeper over one week, being expected to fill out an ESM questionnaire whenever they receive an acoustic signal. This may occur up to eight times per day. While ESM may be applicable in certain research fields, it is rather difficult to apply in studies about video game entertainment. Hence, game researchers make use of different alternatives including questionnaires, electromyograms (EMG), measures of electrodermal activity (EDA), electrocardiograms (ECG) or electroencephalography (EEG) (Nacke, 2009). Due to the limited duration of therapy sessions, such methods should also be used to measure engagement during studies within motor rehabilitation. While measuring engagement directly might be challenging, cognitive engagement could for instance be quantified using questionnaires like the Intrinsic Motivation Inventory (Ryan, 1982), the NASA Task Load Index (Hart and Staveland, 1988, Hart, 2006) or EEG, whereas physical engagement could be assessed using EMG, ECG or EDA.
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1.5 Flow and Augmented Feedback Exercises

Similar to video games, the most frequently mentioned concept that AF exercises inherit, is adjusting the level of difficulty to the capabilities of the patient (Kizony et al., 2004, Reid, 2004, Riva et al., 2006, Kan et al., 2008, Li et al., 2009, Cameirao et al., 2010). However, although the level of difficulty is of great importance, it is not the only characteristic of flow experiences. Hence, I believe, that it should not be the only characteristic to consider, when designing AF exercises for rehabilitation.

The “model of therapeutic engagement” states, that the cognitive willingness of a patient to comply with medical treatment and the behavioral intention to stay engaged during therapy, depends on the perceived treatment benefits (Lequerica and Kortte, 2010). Hence, similar to the flow theory, the goal of an AF exercise should be made apparent to the patient, with an explicit introductory explanation about the benefits that can be achieved before the training starts. In addition, since information about the progress towards reaching a goal is of importance, exercises should further provide frequent explicit feedback about a patient's current level and quality of movement performance.

According to the flow theory, a person should be able to concentrate on the task at hand. Hence, similar to the level of difficulty, the amount of visual stimuli should be chosen in such a way, that patients, e.g., stroke patients with possible perceptual deficits, can concentrate on the goal of the exercise without too much distraction by unnecessary stimuli (see example in Figure 1.1). Thus, the amount and type of visual stimuli should be related to the patient population the exercise is targeted for.

Another component of flow, is “a sense of control over one's activity” (Nakamura and Csikszentmihalyi, 2005). This, however, does not only imply, that the AF exercise should react to the activity of the patient, but also, that the patient knows how to properly interact with the VE. Since the general assumption in rehabilitation is, that motor recovery has to be stimulated by training that aims at the functions to be regained (Dobkin, 2003, Lünenberger et al., 2004), exercises should thus react to functionally meaningful motor performances.
1.5 Flow and Augmented Feedback Exercises

Figure 1.1: AF Exercise Distractions - (a), (b) and (c) illustrate virtual environments with different amount of visual stimuli. While patients might have problems finding animals in (a), removing the trees already drastically simplifies the task. A further simplification can be achieved, by adding elements that indicate the position of the animals in the environment (Image courtesy of Hocoma AG, Switzerland).

Finally, although not directly defined in the flow theory, one should further consider supplementing AF exercises with competitive situations (Sweetser and Wyeth, 2005, Sheehan and Katz, 2012). Competition gives us feedback about our current performance compared to that of others and at the same time encourages us to keep up training, in order to remain competitive. Competition is not only part of our daily life, but in addition can be seen as one of the most prominent mechanics of video games, or rather games in general (Wood et al., 2004).

Thus, in summary I postulate, that AF exercises for motor rehabilitation should

(a) be able to autonomously adapt their level of difficulty to the capabilities of the patient,

(b) provide explicit task goals, making gainable therapy benefits apparent,

(c) provide frequent performance feedback,

(d) allow adjustable visual stimuli,

(e) show functionally meaningful reactions to the motor performance of the patient,

(f) and allow the possibility to accomplish competitive training.
1. Introduction

I believe, that validating these derived characteristics are a first approach to address the lack of guidelines regarding design characteristics for AF exercises.

1.6 Thesis Aims

The literature reviewed in the previous sections makes it apparent that AF exercises are becoming increasingly important in neurological motor rehabilitation. At the same time, the literature further indicates that current studies mostly only assess the deployment effect of AF exercises, instead of addressing the design characteristics that AF exercises should include, in order to optimally enhance engagement and motivation. Since engagement, driven by motivation, is an important factor when it comes to motor learning (Langhorne et al., 1996, Holden, 2005), such guidelines, however, would be of great importance.

The overall aim of this thesis was to assess the effect that AF exercises have on motivation and investigate what characteristics they need to include to increase active participation, i.e., patient engagement, during robot-aided motor rehabilitation. Since, up till now, there are no explicit guidelines of how AF exercises have to be designed, investigations particularly focused on the flow-derived design characteristics set forth in section 1.5.

Since adapting the level of difficulty to the capabilities of the patient is a design characteristic of high importance, I specifically addressed this characteristic in more detail. I hypothesized, that Fitts’ Law (Fitts, 1954) is not only a valid method to describe reaching motions, but that its application, as an adjustment mechanism, will result in AF exercises that are easily adaptable to patients capabilities.

Regarding the other design characteristics, I hypothesized, that AF exercises increase physical patient engagement if they are interactive, i.e., if they provide functionally meaningful reactions to the motor performance of the patients. In addition, AF exercises with explicit task goals and frequent performance feedback increase engagement compared to AF exercises without. Finally, competitive AF exercises increase physical engagement compared to non-competitive exercises.
1.7 Structure

In chapter 2, I assess the general acceptance of AF exercises during robot-assisted gait rehabilitation. While similar studies have already been conducted (König et al., 2008a,b, Wellner et al., 2008, Brütsch et al., 2010, Schuler et al., 2011), the presented results, were among the first to examine the effect of AF exercises on patients, while capturing the observations of therapists. Since the deployment of AF exercises during everyday, clinical therapy mostly depends on the compliance of therapists, their opinion should also be respected and, thus, is of great importance.

Chapter 3 addresses the defined characteristic of adapting the level of difficulty of an AF exercise to the current capabilities of the patient (see section 1.5) for upper-extremity rehabilitation. While several mechanisms have already been proposed for upper-extremity rehabilitation (Cameirao et al., 2010, Kan et al., 2008, Li et al., 2009), most approaches are too complicated to incorporate into different exercises. The mechanism presented in chapter 3, in contrary, can be easily incorporated into a large number of existing and novel, upper-extremity, two-dimensional, pointing AF exercises, independent of their graphical complexity.

In chapter 4 I assess the influence that the defined design characteristics of explicit task goals, frequent performance feedback, functionally meaningful reactions to motor performance and competitive tasks, have on the level of physical engagement, during lower-extremity rehabilitation. Since current AF exercises are not properly analyzed regarding their effectiveness to increase physical engagement during motor rehabilitation (Edmans et al., 2006, Flores et al., 2008) and proposed characteristics for AF exercises that should encourage motivation (Lövquist and Dreifaldt, 2006, Pareto et al., 2008) have not been verified using psycho-physiological measurements, the results presented are a first approach in that direction.

Finally, chapter 5 summarizes the key findings of chapters 2, 3 and 4 and depicts their implications on future developments in the field of AF exercises for motor rehabilitation.
1. Introduction
Augmented Feedback Exercises during Motor Rehabilitation

2.1 Overview

Augmented Feedback (AF) exercises have become increasingly popular in different therapeutic contexts (Hoffman et al., 2000, De Carvalho et al., 2010, Rothbaum et al., 2010, McLay et al., 2012). Likewise, AF exercises have been changing the way patients are motivated to increase engagement during robot-aided motor rehabilitation, enriching the mostly boring and repetitive trainings of long duration. Since the deployment of AF exercises during everyday, clinical therapy mostly depends on the compliance of therapists, feedback about their opinion is of great importance. Hence, the aim of the study in this chapter was, not only to assess the acceptance of this new technology by patients during rehabilitation but also capture the observational feedback of their therapists. Since adapting the level of difficulty to the capabilities of the patient is of great importance, a secondary aim of this study was also to assess how challenging it is for therapists to manually adjust the task difficulty of AF exercises. I hypothesized that patients will like this new technology and that the addition of AF exercises will increase their motivation and, thus, engagement to actively participate during gait rehabilitation.

Most of the text and graphics used in this chapter have previously been published in
2. Augmented Feedback Exercises during Motor Rehabilitation

Zimmerli et al. (2009) and are © 2009 IEEE.

2.2 Methods

2.2.1 System Setup

To increase the motivational aspects of lower-extremity gait training, an AF exercise was developed that consists of different virtual environments and tasks. In order to visualize and render these environments, the open source framework Ogre3D (Ogre3D, 2009) was used. Ogre3D provides a set of powerful tools which simplify the creation of VR-based AF exercises. In order to deliver a high spatial resolution and an adequate visual experience, the environments were displayed on a high definition, 42” flat screen monitor that was placed in front of the patient. Communication between the gait orthosis and the AF exercise was implemented via Ethernet.

2.2.2 Robotic Gait Orthosis

The study was performed using the driven gait orthosis Lokomat® (Hocoma AG, Switzerland, see Figure 2.1) (Colombo et al., 2000, 2001). The Lokomat comprises two actuated leg orthoses that are strapped to the legs of the patient, i.e., an exoskeleton, and is used in conjunction with a body weight support system and a treadmill. The hip (Figure 2.1a) and knee (Figure 2.1b) joints of the orthosis are actuated by linear drives that are integrated into the exoskeleton. The Lokomat can be used with varying supportive force, i.e., guidance force (GF). A GF of 100% allows patients to either walk actively, by pushing into the device, or passively by letting the orthosis move their legs. With 0% GF, the Lokomat is fully transparent, i.e., patients do not feel the orthosis but they have to constantly and actively move their legs. Force sensors in each joint measure the interaction torques between the orthosis and the patient. Measured signals are then used to calculate the so-called biofeedback values per step cycle. Biofeedback values characterize the degree of movement effort, i.e., performance or activity of the patient (Banz et al., 2008a).
2.2 Methods

Figure 2.1: Gait Orthosis - Current version of the driven gait orthosis Lokomat® for lower-extremity rehabilitation (Image courtesy of Hocoma AG, Switzerland). During training, patients walk on a treadmill, while a portion of their weight is unloaded using a body weight support system. Two actuated leg orthoses are strapped to the legs of the patient, moving them on a defined trajectory in the sagittal plane. Augmented Feedback exercises are presented through a screen that is located in front of the patient. The hip (a) and knee (b) joints are actuated by linear drives. Force sensors in each joint measure the interaction torques between the orthosis and the patient.
2. Augmented Feedback Exercises during Motor Rehabilitation

2.2.3 Augmented Feedback Exercise

Although VR allows to create unlimited numbers of both fictional and genuine environments, natural habitats were used to convey a sense of familiarity. For the current study, four different environments were developed ranging from green or snow covered, hilly environments to flat plateaus with rivers and trees (see Figure 2.2). Since rehabilitation should imply active participation, navigation and exploration of each environment was controlled by the patients using exocentric avatars, i.e., through a third person perspective. Patients could chose from one out of four available avatars to allow some degree of customization.

In addition to the simple exploration task, therapists were able to add objects to the environments that had to be collected by the patients. Objects ranged from small coins to big treasure chests, snowmen and barrels. Those were randomly distributed in the environment. By adjusting the type and amount of objects placed, task difficulty could be adjusted to the patient’s capabilities. Objects could be collected by navigating the avatar close to them (required distance depended on object size).

![Figure 2.2: Augmented Feedback Exercise](image)

(a) Screenshot of one of the four created virtual environments that resemble natural habitat with trees and hills. (b) Different objects could be randomly distributed in the virtual environment. These objects had to be collected by the patients during the therapy session (© 2009 IEEE).
2.2 Methods

2.2.4 Movement Metaphor

In order to provoke therapeutically desirable movements and show functionally meaningful reactions to the motor performance of the patient, the forces measured in the hip joints of the orthosis were used to control the avatar’s heading direction and movement speed. Hip and knee angles of the orthosis were mapped onto the virtual joints of the avatars, to increase the feeling of virtual representation.

In order to determine the desired heading in the virtual environment, the biofeedback of the hip currently in swing phase was compared to the biofeedback of the contra lateral hip half a stride earlier, i.e., when the hip was at the same position in the gait cycle as the hip of the swing leg is now (Lünenburger et al., 2006, Banz et al., 2008a). This way, asymmetric physical activity of the legs induces turning in the virtual environment. More specifically, turning to the left could be induced by increasing activity of the right leg (trying to make larger steps) and reducing activity of the left leg (trying to make smaller steps), and vice versa for turning to the right. To account for subjects with generally asymmetric movement patterns, i.e., hemiplegic patients, a global baseline was determined while subjects were instructed to walk straight. This baseline was subtracted from the biofeedback values before the comparison of stance and swing leg was performed (see Lünenburger et al., 2006). Hence,

\[ BF_x(t) = T_{hip\; x}(t) \cdot w_{hip}(g(t)) + T_{knee\; x}(t) \cdot w_{knee}(g(t)) \]  \hspace{1cm} (2.1)

with \( BF_x(t) \) being the biofeedback value at time \( t \), \( T_{hip/knee\; x} \) the recorded torque for the corresponding joint, \( x = L \) the left side, \( x = R \) the right side, \( g(t) \) the gait phase (0-100%) for \( t \), and \( w_{hip}(y) \), \( w_{knee}(y) \) the weighting functions for gait phase \( y \) from (0-100%), according to Banz et al. (2008b),

\[ N_x(t) = BF_x(t) - BL_x(g(t)) \]  \hspace{1cm} (2.2)

with \( N_x(t) \) being the normalized biofeedback and \( BL_x(y) \) the baseline which averages \( BF_x(t_0) \) for all \( t_0 \) with \( y = g(t_0) \) from the baseline recording, and finally
2. Augmented Feedback Exercises during Motor Rehabilitation

\[ \Delta \phi(t) = 0.02 \times s(g(t)) \times (-N_L(t) + N_R(t - 0.5 \times gc) + N_R(t) - N_L(t - 0.5 \times gc)) \tag{2.3} \]

where \( \Delta \phi \) is the virtual rotation angle, \( s(g(t)) \) equals +1 during left swing phase, \(-1\) otherwise, \( gc \) the duration of the gait cycle, such that \( t - 0.5gc \) corresponds to the time half a cycle ago, i.e., the corresponding gait phase of the contralateral leg.

In contrast, not the difference but the sum of the activity of both legs was used to determine the virtual movement speed of the avatar in the virtual environment (VE). The mapping of a subject’s physical activity to the virtual speed was done by averaging the biofeedback values of the hip joints during the swing phase of the last step (Lünenburger et al., 2006, Banz et al., 2008a). Averaged biofeedback values below a certain lower biofeedback threshold resulted in a virtual speed of 0 km/h, those above an upper biofeedback threshold in a virtual speed of 10 km/h. Averaged values between the thresholds were interpolated linearly. Hence,

\[ v_{\text{virtual}} = \begin{cases} \frac{BF_{\text{average}} - Th_{\text{lower}}}{Th_{\text{upper}} - Th_{\text{lower}}} \times 10 \text{ km/h} & \text{otherwise} \\ 0 \text{ km/h} & \text{for } BF_{\text{average}} \leq Th_{\text{lower}} \\ 10 \text{ km/h} & \text{for } BF_{\text{average}} \geq Th_{\text{upper}} \end{cases} \tag{2.4} \]

with \( Th_{\text{lower}} \) and \( Th_{\text{upper}} \) being the lower/upper biofeedback thresholds, \( v_{\text{virtual}} \) the virtual speed and

\[ BF_{\text{Averaged}} = \text{mean}(BF_{\text{Right Hip Swing}}, BF_{\text{Left Hip Swing}}) \tag{2.5} \]

where \( BF_{\text{Right Hip Swing}} \) and \( BF_{\text{Left Hip Swing}} \) are the measured biofeedback values during the swing phase for the right and left hip respectively.

Lower and upper biofeedback thresholds could be adjusted by the therapist during the course of the training. Hence, if the overall activity of the subject increased (e.g., by trying to make larger steps symmetrically on both sides), the movement speed of the avatar increased as well.

Thus, movement speed and heading of the avatar could be controlled independently
by the patient.

### 2.2.5 Subjects

19 subjects aged 16 to 83 (mean age: 52.63, SD: 22.72) participated in the study. Half of them had neurological deficits resulting from stroke (N=11), whereas the rest had craniocerebral injuries (N=4), cerebral hemorrhage (N=2) and spinal cord injuries (N=2). Exclusion criteria were severe contractures or skin lesions in lower limbs, osteoporosis, cardiovascular instability, uncontrolled spasticity that would significantly interfere with the movement of lower extremities, acute medical illness, taller than 190 cm or heavier than 135 kg. Since this was a feasibility study and no medical or physical measurements were conducted, no ethics approval was obtained.

### 2.2.6 Study Design

The study consisted of a systematic survey questionnaire for both patient and therapist, that had to be filled out after the patient’s first training session with the exercise. The questionnaire consisted of 19 questions. Ten questions were directed towards the patient and covered aspects of motivation, control, satisfaction, stability and usefulness of the AF exercise (see Tables 2.1 and 2.2). Additionally, patients were asked about their opinion of the graphical representation of the environments, avatars and tasks. Responses to questions 1-6 for patients and questions 1-7 & 9 for therapists had to be given on a visual analog scale.

#### Table 2.1: Patient Questions

- List of questions that patients were asked after using the AF exercise.

<table>
<thead>
<tr>
<th>Question #</th>
<th>Question Phrasing</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>Did you like it?</td>
</tr>
<tr>
<td>2</td>
<td>Would you like to use the exercise for your next training?</td>
</tr>
<tr>
<td>3</td>
<td>Did you exert yourself more using the AF exercise?</td>
</tr>
<tr>
<td>4</td>
<td>Does such an exercise make sense?</td>
</tr>
<tr>
<td>5</td>
<td>Was the task difficult?</td>
</tr>
<tr>
<td>6</td>
<td>Did the avatar do what you intended?</td>
</tr>
<tr>
<td>7</td>
<td>What did you like most about the graphics?</td>
</tr>
<tr>
<td>8</td>
<td>What did you like most regarding the avatar?</td>
</tr>
<tr>
<td>9</td>
<td>What did you like most about the task?</td>
</tr>
<tr>
<td>10</td>
<td>What would you change?</td>
</tr>
</tbody>
</table>

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2. Augmented Feedback Exercises during Motor Rehabilitation

<table>
<thead>
<tr>
<th>Question #</th>
<th>Question Phrasing</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>Was the patient able to control the avatar?</td>
</tr>
<tr>
<td>2</td>
<td>Was the patient motivated?</td>
</tr>
<tr>
<td>3</td>
<td>Did the patient try things?</td>
</tr>
<tr>
<td>4</td>
<td>Where you able to adjust the exercise to the patient?</td>
</tr>
<tr>
<td>5</td>
<td>Did the exercise increase patient engagement?</td>
</tr>
<tr>
<td>6</td>
<td>Did the patient feel uncomfortable in the Lokomat?</td>
</tr>
<tr>
<td>7</td>
<td>If yes, were you able to reduce this using AF?</td>
</tr>
<tr>
<td>8</td>
<td>How much time did you need to start the system?</td>
</tr>
<tr>
<td>9</td>
<td>How stable was the system?</td>
</tr>
</tbody>
</table>

2.3 Results

The mean duration of a therapy session was 37 min \((mean: 37.16, SD: 7.00 \text{ min})\). The treadmill speed was on average set to 1.8 km/h \((mean: 1.79, SD: 0.27 \text{ km/h})\). For the collection task an average of 75 objects \((mean: 75.38, SD: 67.53)\) were initially placed in the environment.

2.3.1 Feedback from Patients

Results received from patients showed a high agreement regarding questions 1-4, with medians being 96.15%, 96.15%, 97.78% and 95.56% respectively. Lower agreement was obtained for questions 5 and 6, with medians being 46.67% and 57.78% respectively (see Figure 2.3). Since answers to questions 7-10 were subjective, they can be found in the discussion part of this chapter (see 2.4.2).

2.3.2 Feedback from Therapists

Results received from therapists showed a high agreement regarding questions 2-5, with medians being 96.15%, 96.15%, 97.12% and 97.12% respectively. A lower agreement was obtained for question 1 with a median of 65.39%. (see Figure 2.4). While feedback regarding question 3 resulted in a high median, the distribution was
Figure 2.3: Feedback Patients - Box-Plot representation of the results obtained by the patients. Bottom and top edges of illustrated boxplots indicate the 25th and 75th percentile respectively. Whiskers extend to 1.5 times the height of the box or, to the minimum or maximum values if no case has a value in that range. Outliers (o) have values that do not fall within the whiskers and are marked with an asterisk if their values are more than three times the height of the boxes different from the median, i.e., extreme outliers. Questions are shown in Table 2.1.
2. Augmented Feedback Exercises during Motor Rehabilitation

rather high with the 25th percentile being at 23.08%. Since questions 6-9 were targeted towards the stability and operation of the AF system itself, they were excluded from further analysis.

![Box-Plot representation of the results obtained by the therapists. Questions are shown in Table 2.2.]

**Figure 2.4: Feedback Therapists** - Box-Plot representation of the results obtained by the therapists. Questions are shown in Table 2.2.

### 2.4 Discussion

The aim of this study was to create an AF exercise in order to increase motivation during gait therapy with a driven gait orthosis. Patients liked this new way of feedback and according to their therapists were motivated using it. Nearly all patients wanted to use the exercise again in subsequent sessions and most of them believed, that such an exercise makes sense during therapy. According to the feedback from the patients, the exercise even caused an increase in their activity compared to other trainings.
This effect was also perceived by the therapists. Hence, these responses confirm the stated hypothesis. Interesting was that while the therapist responded that they were able to adjust the level of difficulty to the patient, half of the patients reported, that the task was rather difficult for them.

2.4.1 Movement Feedback

Unlike this high agreement towards the usefulness of such an exercise, patients reported having problems controlling the movement of the avatar. This could have been due to the fact, that the way of turning was different in the virtual environment compared to real walking. One first had to get familiar with it. Furthermore, while patients were fixed in the orthosis and could only walk straight, their avatars were able to change directions in the virtual environment. This could have caused a further mismatch in motor action and sensory perception and thus, may have required additional adaptations by the patients. These problems could also have caused patients to rate the task as difficult.

2.4.2 Graphics

When patients were asked how they like the graphics of the exercise, their answers were quite different. Replies spanned from “animating”, “funny”, “entertaining” or “appealing” to “fine”, “ok”, “antique”, “boring” or “ugly”. Younger people tended to give lower ratings. Most probably because they are used to high-end graphics found in current video games. The graphics however were rather simple. The large variation of responses could also have resulted from the repertory of different environments patients could choose from. Since the main aim, however, was the development of entertaining tasks that are supplementary to rehabilitation, the graphics in the current study only played a secondary role.
2. Augmented Feedback Exercises during Motor Rehabilitation

2.4.3 Study Limitations

The main limitation of this study was the broad range of age and neurological deficits of participants. Especially, the difference in age might have had an effect on patients’ judgments. This was for instance visible in the answers to the questions regarding the liking of the deployed graphics. In general, the answers were, however, consistent and having this broad range of patients helped to receive an overall feedback about the use of augmented feedback exercises during robot-aided motor rehabilitation.

In addition, feedback was obtained through a self-defined questionnaire. Hence, the study may have profited from the use of a more standardized questionnaire like, e.g., the intrinsic motivation inventory (IMI) (Ryan, 1982).

2.5 Conclusion

Results demonstrated the effect that AF exercises have, in increasing the overall engagement of patients to actively participate during gait rehabilitation. Although not all patients were immediately able to control the movements of the avatar, most of them expressed, that they would like to keep the exercise for future training sessions.

The study illustrates the broad range of demands patients have towards the visual stimuli of the VE. Hence, similar to non-clinical video games (see section 1.3), it is crucial for future AF exercises to take such preferences and expectations into consideration.

One possible solution would be to adapt the VE to the current physiological state of the patient using a bio-cooperative approach, i.e., a human-in-the-loop structure, that allows to optimize VE to the mechanical or mental engagement of the patient (Riener et al., 2009). Another possibility to make VEs more engaging could be, to deploy the redefined notion of environmental presence by Zimmerli et al. (Zimmerli and Verschure, 2007), i.e., creating VEs that are capable of dynamically producing ongoing, evolving sensory stimuli, consistent with stimulations one would perceive in the real world, while allowing reciprocal interactions.
2.6 Contribution

All methods presented in this chapter were developed during my thesis. Questionnaires were conducted by Andreas Mayr from the “Department of Neurology, Hospital Hochzirl, Zirl, Austria”.

With the current study, I was able to demonstrate that both patients and therapists consider AF exercises as a valuable amendment for lower-extremity gait rehabilitation. The key finding of the current study, however, based on the responses from the patients to questions 5 and from the therapists to question 4, was that it is challenging to manually adjust the task difficulty of an AF exercise to the capabilities of the patient. Since, however, people only engage in an activity, if the outcome matches the effort at which they perform (Hill et al., 1985, Kukla, 1972, Wright, 1982), adapting the level of difficulty is, of great importance. Hence, the next chapter will focus on a method to autonomously adjust the task difficulty of an AF exercise.
2. Augmented Feedback Exercises during Motor Rehabilitation
3

Adapting Augmented Feedback Exercises to Patient Capabilities

3.1 Overview

In the previous chapter I was able to show that patients and therapists considered augmented feedback (AF) exercises to be a valuable part of motor rehabilitation. At the same time, results showed that, while adapting the level of difficulty to the capabilities of the patient is of great importance, this is rather challenging when doing it manually. Hence, AF exercises should be capable of autonomously adapting the level of difficulty. This, however, can only be achieved if an AF exercise considers the cognitive and physical deficits of patients and incorporates a mechanism that is capable of balancing the difficulty of an exercise, i.e., adapt the difficulty to the current capabilities of the patients (see section 1.5). This eventually may also influence active participation by increasing cognitive engagement.

Several studies have already addressed the need for balancing mechanisms in AF exercises for upper-extremity rehabilitation (Cameirao et al., 2010, Kan et al., 2008, Li et al., 2009), lower-extremity rehabilitation (Kö nig et al., 2011a), as well as in specifically developed robotic devices (Colombo et al., 2007, Krebs et al., 2003). While the proposed approaches have shown to be functional, they require the evaluation of the effects that different VR elements have on the level of difficulty (e.g., object size or
3. Adapting Augmented Feedback Exercises to Patient Capabilities

speed), or the processing of various psycho-physiological measurements. This can be a complex and enduring process, that has to be repeated for each particular exercise and hence complicates the adaptation of these mechanisms to different exercises.

Therefore, I propose a different mechanism for rehabilitation that I believe is capable of adapting the difficulty of an exercise. It is based upon a well established empirical formula, Fitts’ Law (Fitts, 1954), that natively already incorporates and describes different parameters and their effects on the level of difficulty. Since Fitts’ Law is a physiologically valid and widely used descriptor of reaching motions, it can be applied to a variety of different upper-extremity exercises. It has successfully been shown, that Fitts’ Law, which originated from one-dimensional, real-world observations, can also be applied to computer screens and input devices (MacKenzie et al., 1991, MacKenzie, 1992, Soukoreff and MacKenzie, 2004) and is valid in the two dimensional space (MacKenzie and Buxton, 1992, Mottet et al., 1994).


3.1.1 Fitts’ Law

Fitts’ Law describes a mathematical relationship between the time needed to move from a start to a target location and the properties these locations possess. The Shannon formulation of Fitts’ Law (MacKenzie, 1992) is

\[
T = a + b \log_2(1 + \frac{D}{W})
\]

(3.1)

with \(T\) being movement time, \(W\) the size of the target location and \(D\) the distance between the start and the target locations. \(a\) and \(b\) are empirical constants determined through linear regression. The logarithmic term of equation 3.1 has been termed as “Index of Difficulty” (ID):
3.2 Methods

\[ ID = \log_2(1 + \frac{D}{W}) \]  \hfill (3.2)

The relationship between \( T \) and \( ID \) can be obtained by measuring movement times for a number of different \( ID \)s and performing a linear regression on the acquired data. This will yield values for the empirical constants \( a \) and \( b \) which are both distinctive for a person's or patient's current performance and the input device that was used.

Since upper-extremity rehabilitation of stroke patients is concerned with the reacquisition of reaching motions, Fitts' Law allows to assess the current quality of a patient's movement, thus, yielding information about his/her capabilities (Christe et al., 2007, McCrea and Eng, 2005, Smits-Engelsman et al., 2007). This information could hence also be used by a mechanism to adjust the difficulty of an exercise.

The main aim of the study in this chapter, thus, was to verify and validate the use of Fitts’ Law as a difficulty adjustment mechanism in upper-extremity rehabilitation.

### 3.2 Methods

#### 3.2.1 Weight compensation system

The study was performed using a passive 5 degrees of freedom weight compensation system (Armeo\(^\circledR\) Spring, Hocoma AG, Switzerland, see Figure 3.1, commercial version of T-WREX; see Sanchez et al. 2004, Housman et al. 2009). Through integrated springs, the orthosis counterbalances the patient's paretic arm against gravity. This enhances any residual functions of the patient, enabling the training of active reaching movements. At the tip of the exoskeleton, the device includes a pressure sensitive handgrip that can trace hand grip force. Electronic sensors that measure movement and grip force allow patients to interact with AF exercises. In addition to physically supporting the patient's arm, the device further allows measuring the range of motion of each patient to allow full interactions with the virtual workspace.
3. Adapting Augmented Feedback Exercises to Patient Capabilities

Figure 3.1: The experimental setup - Experimental setup using a passive device for upper-extremity rehabilitation (Armeo® Spring, Image courtesy of Hocoma AG, Switzerland).
3.2 Methods

3.2.2 Augmented Feedback Exercise

The exercise that was developed to evaluate the balancing mechanism was divided into two phases. In the first phase, the “assessment phase”, Fitts’ Law was used to assess the patient’s performance. The empirical constants $a$ and $b$ that were gained through this assessment were then used during the second phase, the “exercise phase”, to estimate a patient’s movement times between randomly chosen start and target locations of different sizes, i.e., different $ID$s.

Assessment and exercise phase were both performed using the same AF exercise and task. The only difference was, that while there was a time restriction during the exercise phase, this was not the case during the assessment phase. In order to interact with the AF exercise, real world movements in the frontal plane were mapped onto the mouse position on the screen, giving patients continuous spatial position feedback. Each movement from start to target location was defined as a trial, and could be subdivided into three parts. The “initiation part”, i.e., the starting of the movement, the “movement part”, i.e., the movement between start and target location, and the “closure part”, i.e., the trial completion that patients performed by closing their hand when they reached the target.

For each trial, the patients had to move the cursor to a white circle on the screen that indicated the start location. In order to correctly initiate the timer that measured the movement time, they had to wait there for one second before moving towards the target location. The movement initiation trigger was given by a change in color of the target location from dark to light green accompanied by an auditory cue. Unlike during the “assessment phase”, where no timing information was provided, patients received a visual feedback during the “exercise phase” indicating the time left to reach the target through a gradual change in color of the target location from green to red. A rewarding auditory cue was played if the trial was completed on time, indicating a successful trial, while a unrewarding auditory cue was presented if the target was reached too late, indicating an unsuccessful trial (see Figure 3.2).
3. Adapting Augmented Feedback Exercises to Patient Capabilities

3.2.3 Subjects

Ten patients with hemiparesis, due to stroke, in the subacute phase were enrolled in the study at the Schoen Klinik Bad Aibling, Germany. All met the following inclusion criteria: age between 23 and 73 years (mean age: 50.70, SD: 18.36); hemiparesis after first-time, unilateral stroke; time between lesion and enrollment into the study between 3 weeks and 6 months. Exclusion criteria were: other neurological disorders (e.g., Parkinson’s disease, diabetic polyneuropathy), severe aphasia or dementia (not able to understand the informed consent). In order to obtain an overview of the cognitive abilities and motor functions of the subjects, their physical skills were assessed using the upper extremity section of the Fugl-Meyer Assessment (FMA; mean: 39.40, SD: 10.71) (Fugl-Meyer et al., 1975) and the modified Ashworth scale (Bohannon and Smith, 1987) while their cognitive functions were measured using the Addenbrookes cognitive examination-revised (ACE-R; mean: 78.44, SD:15.15), with a focus on visual-spatial deficit perception (ACE-R visuo-spatial subscore; mean: 14.2, SD:2.1) (Mathuranath et al., 2000). To identify the handedness of patients in daily
3.2 Methods

Table 3.1: List of patients that participated in the study - TSS: Time since stroke [months], Arm: Paretic Arm, m: male, f: female, MCA: Middle cerebral artery, WS: watershed stroke, BG: Basal Ganglia, ACA: Anterior cerebral artery, FMA: Upper-Extremity section of the Fugl-Meyer Assessment, Ashworth: Modified Ashworth Scale (Order: shoulder flexors, shoulder extensors, elbow flexors, elbow extensors, wrist flexors, wrist extensors), ACE-R: Addenbrookes cognitive examination-revised (visuo-spatial subscore), EHI: Edinburgh handedness inventory

<table>
<thead>
<tr>
<th>#</th>
<th>Sex</th>
<th>Age</th>
<th>TSS</th>
<th>Arm</th>
<th>Lesion Location &amp; Type</th>
<th>FMA</th>
<th>Ashworth</th>
<th>ACE-R</th>
<th>EHI</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>m</td>
<td>55</td>
<td>3.7</td>
<td>left</td>
<td>MCA / Ischemic</td>
<td>43</td>
<td>1:0;2:1;0:0</td>
<td>52 (10)</td>
<td>right</td>
</tr>
<tr>
<td>2</td>
<td>m</td>
<td>68</td>
<td>2.5</td>
<td>left</td>
<td>WS / Ischemic</td>
<td>48</td>
<td>0:0;0:0:0:0</td>
<td>-</td>
<td>right</td>
</tr>
<tr>
<td>3</td>
<td>m</td>
<td>37</td>
<td>4.3</td>
<td>right</td>
<td>MCA / Ischemic</td>
<td>48</td>
<td>0:0;1+;0:1:0</td>
<td>89 (16)</td>
<td>right</td>
</tr>
<tr>
<td>4</td>
<td>f</td>
<td>73</td>
<td>2.7</td>
<td>left</td>
<td>BG / Hemorrhagic</td>
<td>37</td>
<td>0:1:0;1:0:0</td>
<td>64 (13)</td>
<td>right</td>
</tr>
<tr>
<td>5</td>
<td>m</td>
<td>38</td>
<td>1.6</td>
<td>left</td>
<td>MCA / Ischemic</td>
<td>54</td>
<td>0:0;0:0:0:0</td>
<td>74 (15)</td>
<td>right</td>
</tr>
<tr>
<td>6</td>
<td>m</td>
<td>56</td>
<td>2.5</td>
<td>right</td>
<td>BG / Hemorrhagic</td>
<td>26</td>
<td>0:0;1+;0:1+</td>
<td>67 (12)</td>
<td>right</td>
</tr>
<tr>
<td>7</td>
<td>f</td>
<td>69</td>
<td>2.0</td>
<td>left</td>
<td>MCA / Ischemic</td>
<td>31</td>
<td>0:1:0;1:1:0</td>
<td>91 (16)</td>
<td>right</td>
</tr>
<tr>
<td>8</td>
<td>m</td>
<td>23</td>
<td>2.0</td>
<td>right</td>
<td>MCA / Ischemic</td>
<td>30</td>
<td>0:0;1:0;2:0</td>
<td>84 (16)</td>
<td>left</td>
</tr>
<tr>
<td>9</td>
<td>m</td>
<td>26</td>
<td>1.6</td>
<td>left</td>
<td>MCA, ACA / Ischemic</td>
<td>51</td>
<td>1:1;1:1+;0</td>
<td>92 (15)</td>
<td>right</td>
</tr>
<tr>
<td>10</td>
<td>m</td>
<td>62</td>
<td>2.0</td>
<td>right</td>
<td>Pons / Ischemic</td>
<td>26</td>
<td>0:0;1:0;+1:0</td>
<td>94 (15)</td>
<td>right</td>
</tr>
</tbody>
</table>

activities on a quantitative scale, the Edinburgh handedness inventory (EHI) was employed (Oldfield, 1971). Assessment results are summarized in Table 3.1. Approval was obtained from the Bavarian State Board of Physicians. All patients or their legal representatives gave their written informed consent before participating in the study.

3.2.4 Study Design

The study design comprised four different conditions (see Figure 3.3): The assessment phase was used to measure 50 different movement times (5 IDs x 10 trials) in order to calculate the empirical constants $a$ and $b$ of Fitts’ Law. Since $a$ and $b$ were needed by the balancing mechanism to estimate the movement times, the assessment phase was always at the beginning of an experiment. The diameter of the start location was set to $60$ px ($1 \text{ px} = 0.026 \text{ cm}$). The distance ($D$) between the start and target location was randomly chosen between $90 - 400$ px. The diameter ($W$) of the target location were then calculated by solving the logarithmic term of equation 3.1 for $W$

\[ W = \frac{D}{2^{1D - 1}} \]

(3.3)

where $ID$ was randomly chosen to be 1.4, 1.6, 1.8, 2.0 or 2.2.
3. Adapting Augmented Feedback Exercises to Patient Capabilities

After the assessment, three conditions with different difficulty levels were randomly presented to the patients. In the easy condition the patient was given double the estimated movement time to reach the target (Equation 3.4), in the balanced condition the available time was equal to the estimated time (Equation 3.5). In the difficult condition, patients had half the time to complete the movement (Equation 3.6). Hence,

$$T_{Easy} = 2 \times T_{Estimated}$$  \hspace{1cm} (3.4)$$

$$T_{Balanced} = T_{Estimated}$$  \hspace{1cm} (3.5)$$

$$T_{Difficult} = \frac{T_{Estimated}}{2}$$  \hspace{1cm} (3.6)$$

with $T_{Easy}$, $T_{Balanced}$ and $T_{Difficult}$ being the available times per condition to reach the target and $T_{Estimated}$ being the estimated movement time, based on the assessment phase, to reach the target.

The easy and the difficult conditions were added to the study design in order to evaluate the balancing mechanism. This allowed direct comparisons between the performance during the different conditions and motivational aspects of the balanced condition to those of the easy and difficult conditions. Easy and difficult conditions were similarly based on the results of the “assessment phase” in order to make their level of difficulty relative to the capabilities of each patient. I believe that this resulted in more objective outcomes than if they would have been preset and equal for each subject. While a fixed number of trials were measured during the “assessment phase”, the easy, difficult and balanced conditions were randomly presented for 2 minutes each. To allow for rest, a 30 seconds resting period (RP) followed each condition, during which the NASA Task Load Index questionnaire was deployed, to measure a patient’s subjective feeling of condition difficulty (Hart and Staveland, 1988, Hart, 2006). In order to obtain a better understanding of the balancing mechanism and to verify its applicability, the experiment was carried out twice per patient, once for the paretic and once for the non-paretic arm. The paretic arm was always measured first.
3.2 Methods

Figure 3.3: Study Design - The study was composed of an assessment (1), an easy (2), balanced (3) and difficult (4) condition. In the easy condition, the patient had twice the amount of the estimated time for the movement. In the balanced condition the available time for the movement was equal to the estimated time and in the difficult condition the patient had half the estimated time for the movement. Each condition was followed by a rest period (RP) during which the NASA Task Load Index questionnaire was deployed to measure subjective feeling of condition difficulty. While the assessment condition was always presented at the beginning of the experiment and comprised 50 trials, conditions 2-4 were randomly presented for 2 minutes each.

3.2.5 Movement Measurements

Subdividing each trial into different parts allowed measuring different functional aspects of reaching movements. Reaction time was measured by the lapse of time between the presentation of the movement cue and the patient leaving the starting location, i.e., start circle. The “movement part” was used to measure the movement trajectory in order to calculate the hand-path ratio (HPR) and record the overall speed of the movement. HPR’s are calculated by dividing the length of the movement trajectory by the length of the most direct path between the start and target locations (Figure 3.4), e.g., a HPR of two would mean that the performed trajectory was twice as long as the straight line from start to target. Effective movement speed was obtained by dividing the distance between the start and target location by the total time of the “movement part”. Eventually, the hand-closing time was measured during the “closure part” of the trial, i.e., the elapsed time between the patient entering the target location and closing his/her hand.
3. Adapting Augmented Feedback Exercises to Patient Capabilities

Figure 3.4: Movement Trajectories - (a) and (b) illustrate movement trajectories between start (+) and target (o) locations for the paretic and non-paretic arm of the same subject respectively. One can clearly see the difference between the two movements. Hence, while the movement of the paretic arm is imprecise and uncontrolled, this is not the case for the non-paretic arm.

3.2.6 Questionnaire

During the resting period, patients were asked to answer the six questions of the NASA Task Load Index (NASA-TLX; Table 3.2) (Hart and Staveland, 1988, Hart, 2006) in order to assess their subjective measure of condition difficulty. Each question had to be answered using scores from 0 (very low) up to 21 (very high).

Table 3.2: Questions of the NASA-TLX - Used to obtain the subjective workload of a subject (Hart and Staveland, 1988, Hart, 2006).

<table>
<thead>
<tr>
<th>Sub-Scales</th>
<th>Endpoints</th>
<th>Question</th>
</tr>
</thead>
<tbody>
<tr>
<td>Mental Demand</td>
<td>Low/High</td>
<td>How mentally demanding was the task?</td>
</tr>
<tr>
<td>Physical Demand</td>
<td>Low/High</td>
<td>How physically demanding was the task?</td>
</tr>
<tr>
<td>Temporal Demand</td>
<td>Low/High</td>
<td>How hurried or rushed was the pace of the task?</td>
</tr>
<tr>
<td>Performance</td>
<td>Good/Poor</td>
<td>How successful were you in accomplishing what you were asked to do?</td>
</tr>
<tr>
<td>Effort</td>
<td>Low/High</td>
<td>How hard did you have to work to accomplish your level of performance?</td>
</tr>
<tr>
<td>Frustration Level</td>
<td>Low/High</td>
<td>How insecure, discouraged, irritated, stressed and annoyed were you?</td>
</tr>
</tbody>
</table>
3.3 Results

3.2.7 Data Acquisition & Analysis

Experiments were conducted on a commercial PC with a screen resolution of 1280x1024 pixels using the weight compensation device as mouse input. All statistical analyses and plots were generated using IBM SPSS 19 (IBM Corporation, USA) and MATLAB R2009b (MathWorks, USA) on an Intel MacBook Pro. Parameters were compared using a two-way analysis of variance (ANOVA). Post hoc analyses were performed using a Tukey HSD. The significance level was set to $p < 0.05$. For further clarification, the effect size ($\eta^2$) and the results of the F-test with degrees of freedom are indicated.

3.3 Results

3.3.1 Fitts’ Law

The available time for the movements during the easy, balanced and difficult conditions was based on the constants $a$ and $b$ that were calculated during the “assessment” phase (Table 3.3). The squares of the correlation coefficients ($R^2$) between modeled and observed data is at $R^2 > 0.7$ for the paretic arm of five out of ten patients (Patient ID’s 1, 2, 3, 9 and 10) and for the non-paretic arm of six out of ten patients (Patient ID’s 4, 5, 6, 7, 8 and 10). Further analysis of the data did not show any correlation of the constants $a$ and $b$ to the various clinical assessments, e.g. FMA.

Table 3.3: List of empirically determined constants and squares of the correlation coefficients - Obtained for the paretic and non-paretic arm using linear regression in the assessment condition.

<table>
<thead>
<tr>
<th>Patient ID</th>
<th>Paretic Arm</th>
<th>Non-Paretic Arm</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>$a$</td>
<td>$b$</td>
</tr>
<tr>
<td>1</td>
<td>5.2426</td>
<td>-3.8267</td>
</tr>
<tr>
<td>2</td>
<td>1.9382</td>
<td>-1.5806</td>
</tr>
<tr>
<td>3</td>
<td>1.5705</td>
<td>0.1127</td>
</tr>
<tr>
<td>4</td>
<td>2.5553</td>
<td>1.6190</td>
</tr>
<tr>
<td>5</td>
<td>-0.3324</td>
<td>3.3407</td>
</tr>
<tr>
<td>6</td>
<td>0.0606</td>
<td>11.7751</td>
</tr>
<tr>
<td>7</td>
<td>2.4267</td>
<td>0.7160</td>
</tr>
<tr>
<td>8</td>
<td>3.0954</td>
<td>6.7390</td>
</tr>
<tr>
<td>9</td>
<td>0.5810</td>
<td>0.1949</td>
</tr>
<tr>
<td>10</td>
<td>1.7999</td>
<td>-0.8733</td>
</tr>
</tbody>
</table>
3. Adapting Augmented Feedback Exercises to Patient Capabilities

3.3.2 Successful Trials

The medians of the percentage of successful trials for both arms showed a gradual decrease with increasing condition difficulty (Figure 3.5). The ANOVA did not reveal a significant difference between the paretic and non-paretic arm ($F_{1,60} = 3.768; \ p = 0.057; \ \eta^2 = 0.065$), and showed a significant difference between the conditions ($F_{2,60} = 44.623; \ p < 0.01; \ \eta^2 = 0.623$), with no significant interactions between arm and conditions ($F_{1,60} = 0.037; \ p = 0.964; \ \eta^2 = 0.001$). Post hoc analysis revealed that with both the paretic and the non-paretic arm, the difficult condition significantly differed from the easy condition ($p < 0.01$). In the non-paretic arm there was an additional significant difference between the balanced and the difficult condition ($p < 0.01$).

![Figure 3.5: Successful Trials](image)

Figure 3.5: Successful Trials - Boxplots show the decreasing amount of successful trials over condition difficulty for the paretic and non-paretic arm. Bottom and top edges of illustrated boxplots indicate the 25th and 75th percentile respectively. Whiskers extend to 1.5 times the height of the box or, to the minimum or maximum values if no case has a value in that range. Outliers (o) have values that do not fall within the whiskers and are marked with an asterisk if their values are more than three times the height of the boxes different from the median, i.e., extreme outliers.
3.3 Results

3.3.3 Movement Speed

Reducing the available time to reach the target resulted in faster movement speeds (Figure 3.6). The ANOVA revealed a significant difference between the paretic and non-paretic arm ($F_{1;59} = 19.346; p < 0.01; \eta^2 = 0.267$), and showed a significant difference between the conditions ($F_{2;59} = 6.013; p < 0.01; \eta^2 = 0.185$), with no significant interactions between arm and conditions ($F_{2;59} = 1.572; p = 0.217; \eta^2 = 0.056$). Post hoc analysis revealed a significant difference between the paretic and non-paretic arm for the difficult condition ($p < 0.01$). Significant differences were also found between the easy and difficult conditions ($p = 0.016$) and the balanced and difficult conditions ($p = 0.033$) in the non-paretic arm.

![Figure 3.6: Movement Speed](image)

**Figure 3.6: Movement Speed** - Boxplots illustrate the movement speeds in m/s of the three conditions easy, balanced and difficult for the paretic and non-paretic arm.

3.3.4 Hand-Closing Time

The hand-closing time decreased with decreasing available time (Figure 3.7). While the ANOVA showed a significant difference between the paretic and non-paretic arm ($F_{1;59} = 7.672; p < 0.01; \eta^2 = 0.126$), no significant differences were found between
3. Adapting Augmented Feedback Exercises to Patient Capabilities

the different conditions ($F_{2;59} = 2.620; \ p = 0.082; \ \eta^2 = 0.09$), and the interaction between arm and conditions ($F_{2;59} = 0.806; \ p = 0.452; \ \eta^2 = 0.03$).

![Boxplots showing hand-closing time for different conditions](image)

**Figure 3.7: Hand-Closing Time** - Boxplots showing the hand-closing time for the three different conditions easy, balanced and difficult for the paretic and non-paretic arm.

3.3.5 Reaction Time

Reaction time remained constant over the different conditions (Figure 3.8). The ANOVA did not show any significant difference between the paretic and non-paretic arm ($F_{1;59} = 0.027; \ p = 0.871; \ \eta^2 = 0.001$), no significant differences between the different conditions ($F_{2;59} = 0.978; \ p = 0.383; \ \eta^2 = 0.036$), and no significant differences for the interaction between arm and conditions ($F_{2;59} = 0.423; \ p = 0.657; \ \eta^2 = 0.016$).
3.3 Results

Figure 3.8: Reaction Times - Boxplots illustrating the reaction times.

3.3.6 Hand-Path Ratio

The ANOVA of the hand-path ratio showed a significant difference between the paretic and non-paretic arm (Figure 3.9; $F_{1;59} = 9.357; p < 0.01; \eta^2 = 0.150$), no significant differences were found between the different conditions ($F_{2;59} = 0.054; p = 0.947; \eta^2 = 0.002$), and the interaction between arm and conditions ($F_{2;59} = 0.014; p = 0.986; \eta^2 = 0.001$).
3. Adapting Augmented Feedback Exercises to Patient Capabilities

Figure 3.9: Hand-path ratios - Boxplots illustrating the hand-path ratios for the paretic and non-paretic arm.

3.3.7 NASA-TLX

The ANOVA of the NASA-TLX did not reveal any significant differences between the paretic and non-paretic arm for any of the sub-scales (Figure 3.10). Significant differences between conditions were found for the performance sub-scale in the paretic arm between the easy and difficult condition \( (p < 0.01) \), in the non-paretic arm between the easy and difficult conditions \( (p < 0.01) \) as well as in the non-paretic arm between the balanced and difficult \( (p < 0.01) \) conditions. In the effort sub-scale a significant difference was found in the non-paretic arm for the easy and difficult conditions \( (p = 0.04) \).
3.4 Discussion

In the current study I investigated the applicability and validity of a mechanism that should be useful for adapting the difficulty of the exercise to the capabilities of the patient for upper-extremity rehabilitation. This difficulty balancing mechanism is based
3. Adapting Augmented Feedback Exercises to Patient Capabilities

on Fitts’ Law.

The results show a decreasing percentage of successful trials from the easy to the difficult conditions. This seems obvious as patients had less time to reach the target on time and complete the trial. Although the difficulty was adapted separately, the ANOVA did not reveal any significant differences between the paretic and non-paretic arm. I, thus, state that Fitts’ Law is a valid mechanism to balance the difficulty of upper-extremity, two-dimensional, pointing exercises. Hence, although patients are capable of moving their non-paretic arm more accurately and faster, the percentage of successful trials is similar compared to when they perform the movements with their paretic arm.

The developed exercise further allowed assessing different movement characteristics, i.e., movement speed, hand-closing time, reaction time and the hand-path ratio. In contrast to the reaction times, all other measurements showed a significant difference between the paretic and non-paretic arm reflecting the impaired motor control. While this seems obvious, the interesting aspect is that movement speed increased and hand-closing time decreased with increasing condition difficulty while the hand-path ratio stayed the same. Although one could assume that an increase of the movement speed could cause decreased movement accuracy, this was not the case. It rather influenced whether the target was reached or not. Increased speed could, however, be used to train the hand-closing functionality and movement speed of the paretic arm.

3.4.1 Outcome Perception & Performance

Different theories on motivation state that performance is influenced by the effort a person invests during an exercise and the outcome they perceive (Hill et al., 1985, Kukla, 1972, Wright, 1982). The effort, performance and frustration sub-scales of the NASA-TLX visualize this. Although the only significant difference for the paretic arm was found in the performance subscale, the others show a trend that while patients increase their effort with increasing condition difficulty, they perceive their performance as getting worse and get frustrated. This eventually may lead to disengagement. Hence, the level of difficulty may not only affect cognitive but also emotional engagement. Assuming that a success rate of 100% during an exercise may cause
boredom and anything below 50% frustration, using Fitts’ Law to adjust difficulty, gives an initially challenging but at the same time not frustrating exercise difficulty. By manipulating the estimated time one can adjust the difficulty and thus, influence a patient’s awareness of his/her capabilities, thereby increasing engagement during therapy.

3.4.2 Study Limitations

Despite these findings, some limitations of the current study design need to be noted. In the current setup, patients had to complete a trial by closing their hand when they reached the target. While this was useful to gain some insight into patients’ hand-closing functionality, it at the same time limited the number of patients that were able to participate. As an alternative, trial completion could be realized by asking patients to simply move over the target. This would allow inclusion of more patients with the drawback of losing one of the metrics.

In order to acquire the constants $a$ and $b$ for the balancing mechanism, 50 trials were measured during the “assessment phase”. As the results show, the correlations of the linear regressions are quite low in some patients. This probably caused sporadic inaccurate estimates during the different conditions. According to Figure 3.5, these, however, did not have a large effect on the number of successful trials. The number of assessment trials was limited to 50, in order to prevent patients of getting bored by the rather simple task.

Another interesting finding, when looking at the squares of the correlation coefficients, is the difference between the paretic and non-paretic arm. While five out of then patients (Patient ID’s 1, 2, 3, 9 and 10) had high correlations ($R^2 > 0.7$) for the paretic arm, they showed low correlations for the non-paretic arm. Although there seems to be a visible pattern when looking at this distribution, the clinical evaluation does not show any functional differences between the two groups that would allow drawing any conclusions. The sometimes-low correlations for the non-paretic arm where probably due to the fact that patients were not used to train it.

During the “exercise phase” each condition (easy/balanced/difficult) was presented and performance was measured for 2 minutes. Therefore, the number of trials in each
3. Adapting Augmented Feedback Exercises to Patient Capabilities

condition differed. This approach, compared to having a fixed number of trials per condition, was chosen to guarantee equal temporal lengths, preventing exhaustion, which could have negatively affected the performance during subsequent conditions.

In order to obtain a subjective measure of condition difficulty, patients were asked to give feedback using the NASA-TLX. Although few significant differences were found, results show a trend indicating that condition difficulty is followed by an increase of mental demand, effort and frustration levels as well as by a decrease of performance. As I, however, also conclude on the motivation level of the patients, using a more established and motivation-oriented questionnaire, e.g., the intrinsic motivation inventory (IMI) (Ryan, 1982) would have been favored. However, in the current study, the NASA-TLX was chosen over the IMI because of the smaller number of questions patients had to answer. This kept the cognitive demand low during the rest period.

3.4.3 Graphics

Because the aim of the study in this chapter was to verify the applicability of the balancing mechanism, the graphical implementation of the current exercise was kept simple. Due to its simplicity, the mechanism can, however, easily be adapted to exercises of higher graphical complexity. By substituting the background image as well as the illustrations for the start and target location, for instance, the graphical appearance of the current exercise can be made more appealing, while keeping the underlying game mechanics.

3.5 Conclusion

Fitts’ Law provides a powerful mechanism for adapting task difficulty. Because it is physiologically valid and a widely used descriptor of reaching movements, Fitts’ Law is an ideal estimator for upper-extremity rehabilitation, in which patients have to relearn both speed and accuracy of movements. The simplicity of the mechanism further allows incorporating it in a large number of existing and novel, upper-extremity, two-dimensional, pointing AF exercises, independent of the graphical complexity those
exercises exhibit. By varying the distance and size of start and target locations, Fitts’ Law additionally allows an unlimited number of variations and, thus, a high diversification of visual stimuli in AF exercises.

3.6 Contribution

All methods presented in this chapter were developed during my thesis. Patient data was recorded by Carmen Krewer from the “Schoen Klinik Bad Aibling, Bad Aibling, Germany”.

With the current study, I was the first to prove that Fitts’ Law is not only a descriptor of reaching motions in combination with accuracy, but, that it can also be successfully used as a balancing mechanism, to adapt the difficulty of a task to the capabilities of a patient. Incorporating Fitts’ Law into upcoming AF exercises for upper-extremity rehabilitation, may thus increase a patients’ cognitive engagement and motivation over a longer period of time. Based on the results of the previous chapter (2) and the current study, I state, that difficulty adaptation is an important element to consider, when developing guidelines for designing AF exercises.

The effects of the remaining design characteristics, on increasing engagement, that were derived in section 1.5 will be subject in the next chapter.
3. Adapting Augmented Feedback Exercises to Patient Capabilities
4

Increasing Engagement using Augmented Feedback Exercises

4.1 Overview

According to the characteristics set forth in the introduction (see section 1.5), augmented feedback (AF) exercises should (a) be able to adapt their level of difficulty to the capabilities of the patient, (b) provide explicit task goals, making therapy benefits apparent, (c) provide frequent performance feedback, (d) allow adjustable visual stimuli, (e) show functionally meaningful reactions to the motor performance of the patient and (f) allow the possibility to accomplish competitive training.

In the previous chapter, I already was able to successfully validate a mechanism to adapt the level of difficulty of an AF exercise to the capabilities of the patient, i.e., characteristic (a). Hence, the aim of the current study, was to further assess the influence of the remaining characteristics (b-f), except for adjustable visual stimuli (d), on the level of physical engagement during lower-extremity, i.e., gait rehabilitation. Measuring the effects of different visual stimuli on cognitive and emotional engagement is an important, extensive task that should be addressed in more detail, with specific studies.

Thus, in the current study, I hypothesized that (1) AF exercises only increase a patients’ physical engagement if they are interactive, i.e., if they provide functionally meaningful
4. Increasing Engagement using Augmented Feedback Exercises

Increasing reactions to the motor performance of the patients. In addition, (2) AF exercises with explicit task goals and frequent performance feedback increase physical engagement compared to AF exercises without any goals and no performance feedback. Finally, (3) competitive AF exercises increase physical engagement compared to non-competitive exercises.


4.2 Methods

4.2.1 Robotic Gait Orthosis

The study was performed using the driven gait orthosis Lokomat® (Hocoma AG, Switzerland, see Figure 2.1) (Colombo et al., 2000, 2001). The Lokomat comprises two actuated leg orthoses that are strapped to the legs of the patient, i.e., an exoskeleton, and is used in conjunction with a body weight support system and a treadmill. The hip and knee joints of the orthosis are actuated by linear drives that are integrated into the exoskeleton. Force sensors in each joint measure the interaction torques between the orthosis and the patient. Measured signals are then used to calculate the so-called biofeedback values per step cycle. Biofeedback values characterize the degree of movement effort, i.e., performance or activity of the patient (Banz et al., 2008a).

4.2.2 Augmented Feedback Exercises

In order to verify the hypotheses, different AF exercises were created using the Unity3D game engine (Unity3D, 2012) and projected in front of the subjects onto a screen of two meters diameter using a beamer with a resolution of 1280x800 (Sony WXGA-Projector VPL-SW535). Tasks within the exercises were presented in a detailed, virtual environment (VE). Subjects walked along a straight path in a natural, hilly landscape of varying flora. Bird and locust chirps were added as background sounds, to further increase the AF-experience (see Figure 4.1). Four exercises were implemented:
4.2 Methods

**Steady:** Subjects were walking along the path with a constant speed, i.e., no interaction with the virtual environment was possible (see Figure 4.1a).

**Speed:** The activity of the subjects in the orthosis, i.e., their measured biofeedback values, was used to modulate the virtual speed in the environment, as described below in further detail (see sub-section 4.2.3). More active participation resulted in faster, virtual movement speeds (see Figure 4.1b).

**Sprint:** While the mapping of the activity to the virtual speed was the same as in the Speed condition, subjects were additionally informed about their average speeds over the last and second to last 100 meters. Results were presented using numeric values in km/h at the top of the screen, when subjects passed “distance indication signs” that were placed at the side of the path (see Figure 4.1c and d). An attentive audio cue was played, to make subjects aware of this information. This allowed further performance comparisons.

**Race:** The activity of subjects was again mapped onto the virtual speed, similar as in the Speed and Sprint conditions. The Race condition, however, included competition. Subjects had to compete with a virtual opponent walking next to them, in a race-like manner (see Figure 4.1e and f). In order to respect the design characteristic of adjusting the level of difficulty to the subjects’ capabilities, the virtual opponent was programmed in such a way, that it always stayed near the subject. This was achieved, by mapping the subjects’ speed onto the opponent, with a 5 second delay. In addition to the visual feedback of seeing the opponent, game-like, high-frequency success and low-frequency fail audio cues were played, when subjects passed or were passed by the competitor, respectively.

Hence, while in the Steady condition, subjects had no interaction possibilities, the Speed, Sprint and Race conditions allowed interaction through the subjects’ activity in the gait orthosis, providing a “sense of control”. Compared to the Speed condition, the Sprint and Race conditions further introduced challenging tasks with explicit performance goals. Hence, while the goal in the Sprint condition was to exceed the speed of the previous 100 meters, the goal in the Race condition was to outperform the opponent. Finally, the Race condition included competition as an additional element.
4. Increasing Engagement using Augmented Feedback Exercises

Figure 4.1: Augmented Feedback Exercises - Natural, hilly landscape of varying flora with bird and locust chirps in the background. Subfigures (a) and (b) represent the Steady and Speed conditions, respectively. Subfigures (c) and (d) illustrate the audiovisual feedback of the Sprint and subfigures (e) and (f) the audiovisual feedback of the Race condition (Avatar courtesy of Hocoma AG, Switzerland).
4.2 Methods

4.2.3 Virtual Speed Mapping

The mapping of a subject’s physical activity to the virtual speed was done by averaging the biofeedback values (Banz et al., 2008a) of the hip joints during the swing phase of the last step. The swing phase was favored over the stance phase, since it allowed clearer instructions to patients on how to interact with the AF exercise. Averaged biofeedback values below a certain lower threshold resulted in a virtual speed of 0 km/h, those above an upper threshold in a virtual speed of 10 km/h (see Tables 4.1 and 4.2 for min/max threshold values). Averaged values between the thresholds were interpolated linearly. Hence,

\[
v_{\text{virtual}} = \begin{cases} 
\frac{BF_{\text{average}} - Th_{\text{min}}}{Th_{\text{max}} - Th_{\text{min}}} \times 10 \text{ km/h} & \text{otherwise} \\
0 \text{ km/h} & \text{for } BF_{\text{average}} \leq Th_{\text{min}} \\
10 \text{ km/h} & \text{for } BF_{\text{average}} \geq Th_{\text{max}} 
\end{cases}
\]  

(4.1)

with \( Th_{\text{min}} \) and \( Th_{\text{max}} \) being the min/max thresholds, \( v_{\text{virtual}} \) the virtual speed and

\[
BF_{\text{Averaged}} = \text{mean}(BF_{\text{Right Hip Swing}}, BF_{\text{Left Hip Swing}})
\]  

(4.2)

where \( BF_{\text{Right Hip Swing}} \) and \( BF_{\text{Left Hip Swing}} \) are the measured biofeedback values during the swing phase for the right and left hip, respectively.

In control subjects, the thresholds were fixed for males to -800 and 800, for females to -600 and 600. These thresholds were used to quantify the walking activity of the subjects. A value of zero would mean, that a subjects’ walking pattern perfectly matches the current movement of the device. Positive values indicate, that the movement of the subject is corrected by the device in the direction of joint extension, i.e. the subject is actively participating. Negative values indicate, that the device has to correct the movement in the direction of joint flexion, i.e., the subject is passive or doing a wrong movement (Lünenburger et al., 2004, Banz et al., 2008a). Hence, the higher the thresholds, the more physical activity was needed to reach the maximum movement speed.

To adapt the thresholds to the varied capabilities of the subjects with SCI, subjects were asked at the beginning of the experiment, to try to be as active and subsequently...
4. Increasing Engagement using Augmented Feedback Exercises

as passive as possible. The resulting minimal and maximal biofeedback values were then used as min/max thresholds for the course of the experiment. In two subjects with SCI, the thresholds had to be slightly adjusted during the beginning of the experiment (marked by brackets in Table 4.2).

4.2.4 Subjects

Ten control subjects without any neurological movement disorders (see Table 4.1) and twelve subjects with a spinal cord injury (SCI) (see Table 4.2) were enrolled in the study at the Paraplegic Centre of Balgrist University Hospital, Switzerland. Inclusion criteria for the SCI subjects was to be able to stand upright for at least 30 seconds with or without support. Both chronic (>1 year post lesion) and acute (<1 year post lesion) SCI subjects were included. Exclusion criteria were signs of depression, severe contractures or skin lesions in lower limbs, osteoporosis, cardiovascular instability, uncontrolled spasticity that would significantly interfere with the movement of lower extremities, acute medical illness, taller than 190 cm or heavier than 135 kg. Control subjects were aged between 23 and 31 years (mean age: 25.9, SD: 2.73), subjects with SCI between 23 and 66 years (mean age: 46.3, SD: 14.0). Mobility was quantified using the “Walking Index for Spinal Cord Injury II” (WISCI II; mean: 15.4, SD: 6.37) (Ditunno et al., 2000) and the mobility sub-scale of the “Spinal Cord Independence Measure III” (SCIM III; mean: 29.7, SD: 8.48) (Itzkovich et al., 2007). Both WISCI II and SCIM III have been shown to be valid and reliable for patients with spinal cord lesions (Itzkovich et al., 2007, Burns et al., 2011). In the WISCI II, patients are evaluated on a scale from zero to twenty, where zero means standing and walking are impossible and twenty indicates normal walking capabilities (Ditunno et al., 2000). In contrast to manually measuring the WISCI, in the current study, subjects with SCI gave oral feedback. SCIM III covers everyday tasks and considers the economic burden of disability. In this study, I assessed the sub-scale of mobility, where zero means dependent in all areas and forty indicates independency (Itzkovich et al., 2007).

Lower extremity functions were assessed using the “Lower Extremity Motor Score” (LEMS; mean: 35.3, SD: 12.0) (Maynard et al., 1997, Marino and Graves, 2004, Furlan et al., 2008). The LEMS is a subscale of the ASIA motor score, which is a widely
accepted and applied measure to assess muscle strength (Marino et al., 2003). It rates the strength of five key muscles in each leg. While a score of zero indicates total paralysis, five implies normal strength. The overall score is then calculated by adding up all sub-scores.

Finally, the level of depression was measured using the “Beck Depression Inventory II” (BDI II; mean: 7.6, SD: 4.3) (Krefetz et al., 2002). BDI II yields reliable, internally consistent and valid scores that consist of 21 items with each having a four-point scale ranging from 0 to 3. Hence, the maximum total score of the BDI II is 63 (Arnau et al., 2001).

Approval was obtained from the local ethics committee according to the ICH-GCP-guidance. All subjects gave their written informed consent before participating in the study.

Table 4.1: List of control subjects that participated in the study - m: male, f: female, Speed: Treadmill speed in km/h, Min/Max: Biofeedback thresholds. Subjects were asked regarding their dominant leg.

<table>
<thead>
<tr>
<th>#</th>
<th>Gender</th>
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<th>Dominant Leg</th>
<th>Speed</th>
<th>Min / Max</th>
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<td>24</td>
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<td>2.0</td>
<td>-800 / 800</td>
</tr>
<tr>
<td>2</td>
<td>f</td>
<td>24</td>
<td>right</td>
<td>2.0</td>
<td>-800 / 800</td>
</tr>
<tr>
<td>3</td>
<td>m</td>
<td>27</td>
<td>right</td>
<td>2.0</td>
<td>-800 / 800</td>
</tr>
<tr>
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<td>m</td>
<td>26</td>
<td>right</td>
<td>2.0</td>
<td>-800 / 800</td>
</tr>
<tr>
<td>7</td>
<td>f</td>
<td>24</td>
<td>right</td>
<td>2.0</td>
<td>-600 / 600</td>
</tr>
<tr>
<td>8</td>
<td>f</td>
<td>26</td>
<td>right</td>
<td>2.0</td>
<td>-600 / 600</td>
</tr>
<tr>
<td>9</td>
<td>m</td>
<td>31</td>
<td>right</td>
<td>2.0</td>
<td>-800 / 800</td>
</tr>
<tr>
<td>10</td>
<td>m</td>
<td>24</td>
<td>right</td>
<td>2.0</td>
<td>-800 / 800</td>
</tr>
</tbody>
</table>
### Table 4.2: List of subjects with SCI that participated in the study

- m: male, f: female, Lesion: Level of Lesion, C: Cervical, T: Thoracic, AIS: American Spinal Injury Association Impairment Scale; AIS classification: A = sensorimotor complete, B = motor complete, C = sensorimotor incomplete (less than 50% of key muscles below neurological level with muscle grade less than 3 of 5), D = sensorimotor incomplete (more than 50% of key muscles below neurological level with muscle grade 3 or more), TSI: Time Since Injury (years), BWS: Body Weight Support (kg), Speed: Treadmill speed in km/h, Min/Max: Biofeedback thresholds (values in brackets were slightly adjusted during the experiment), SCIM III: Spinal Cord Independence Measure III, WISCI II: Walking Index for Spinal Cord Injury II, LEMS: Lower Extremity Motor Score, BDI II: Beck Depression Inventory II

<table>
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<th>TSI</th>
<th>Etiology</th>
<th>BWS</th>
<th>Speed</th>
<th>Min / Max</th>
<th>SCIM III</th>
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<td>12</td>
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<td>f</td>
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<td>D</td>
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<td>C</td>
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<td>-300 / 300</td>
<td>27</td>
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<td>12</td>
</tr>
<tr>
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<td>40</td>
<td>2.0</td>
<td>-300 / 100</td>
<td>28</td>
<td>16</td>
<td>40</td>
<td>12</td>
</tr>
</tbody>
</table>
4.2 Methods

4.2.5 Study Design

In total, the study design comprised six different conditions (see Figure 4.2). These included the four previously described conditions, i.e., Steady, Speed, Sprint and Race, with the addition of a start and end baseline. While the start and end baselines were always the first and the last conditions, respectively, the four other conditions were presented in a pseudorandom order, i.e., while the Steady and Speed conditions were presented in a random order during the first half, the Sprint and Race conditions were randomly presented during the second half of the experiment. During the baselines, no AF-environment was presented. Each condition was shown for four minutes. At the beginning of each condition, subjects were instructed for 10 seconds, what they should try to achieve during the course of the condition, e.g., “Try to walk actively” or “Modulate the speed using your activity”. To allow for rest between the different conditions, a relaxation period of three minutes followed each condition.

![Figure 4.2: Study Design](image)

**Figure 4.2: Study Design** - The study design comprised six different conditions: the start and end baselines, the Steady, the Speed, the Sprint and the Race condition. Between conditions there were resting periods (RP). Each condition lasted 4min, RP 3min. Conditions where presented in a pseudorandom order, i.e., the Steady and Speed conditions as well as the Sprint and Race conditions were randomly presented.

4.2.6 Measurements

Engagement can be defined as a construct that is driven by motivation and executed through active, effortful participation (Lequerica and Kortte, 2010). Since, in the current study, active participation results in increased muscle activity, I argue that electromyography (EMG) can be used to measure engagement (De Luca, 1997, Disselhorst-Klug et al., 2009). Hence, EMGs were measured in both legs at the tibialis...
4. Increasing Engagement using Augmented Feedback Exercises

anterior, the gastrocnemius medialis, the rectus femoris and biceps femoris (Noraxon TeleMyo™ DTS). In addition, biofeedback values of the gait orthosis at each joint during stance and swing phase were recorded (Banz et al., 2008a). The physical load of the subjects was assessed through electrocardiogram (ECG) accurate heart rate measurements.

4.2.7 Data Acquisition & Analysis

EMG recordings were made from bilateral proximal and distal leg muscles using surface electrodes. The EMG signals were amplified (400 times), filtered (band-pass 10500 Hz) and sampled at 1500 Hz. In addition, left and right heel strikes were recorded. The locomotor EMG signals of all recorded muscles were offset corrected and EMG values were calculated as the mean root mean square (RMS) per condition of stance and swing phase, respectively. Biofeedback values were calculated for each recorded stride (bilateral hip and knee during the stance and swing phases) and averaged per condition. Heart rate measurements were done using “Polar WindLink™” and a “Polar WearLink®” chest strap. Mean heart rate was calculated per condition. Data acquisition and analysis were both conducted on commercial PCs. All statistical analysis and plots were generated using IBM SPSS 19 (IBM Corporation, USA) and MATLAB R2009b (MathWorks, USA). In order to compare the measured EMG results, the data of each subject was normalized to his/her individual start baseline. Since the biofeedback values were used to interact with the virtual environment, they were excluded from analysis. Start and end baseline did not differ significantly. Therefore, the end baseline was removed from data analysis. Due to the non-normal distribution of the data, statistical analysis between all conditions was performed using a non-parametric Friedman test. If results were significant (p < 0.05), post-hoc analysis between paired conditions was done using a Wilcoxon signed rank test with the level of significance corrected according to Holm-Bonferroni (Holm, 1979).
4.3 Results

4.3.1 Heart Rate

Post-hoc analysis for subjects with SCI revealed significant differences between the Steady and the Speed ($p = 0.003$) and between the Steady and the Sprint condition ($p = 0.006$) (see Figure 4.3A). In control subjects, statistical post-hoc analysis showed significant differences between the Steady and the Race ($p = 0.007$) and between the Speed and the Sprint condition ($p = 0.009$) (see Figure 4.3B). One subject of the SCI group had to be removed from analysis because of recording issues with the chest belt.

Figure 4.3: Heart Rate - Heart rate for subjects with SCI (A, N=11) and control subjects (B, N=10), per condition, compared to the start baseline. Bottom and top edges of illustrated boxplots indicate the 25th and 75th percentile, respectively. Whiskers extend to 1.5 times the height of the box or, to the minimum or maximum values if no case has a value in that range. Outliers (o) have values that do not fall within the whiskers and are marked with an asterisk if their values are more than three times the height of the boxes different from the median, i.e., extreme outliers.
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4.3.2 EMG - Biceps Femoris

Post-hoc analysis of the RMS values for the biceps femoris in subjects with SCI showed significant differences in the right leg during swing phase between the Steady and the Speed (\(p = 0.013\)), the Steady and the Sprint (\(p = 0.003\)), the Steady and the Race (\(p = 0.004\)) and between the Speed and the Sprint condition (\(p = 0.006\)) (see Figure 4.4A). In control subjects, the only significant difference was found in the left leg during stance phase between the Steady and the Race condition (\(p = 0.007\)) (see Figure 4.4B). One subject with SCI had to be removed from the analysis of the right leg because of a loosened EMG electrode.

**Figure 4.4: Biceps Femoris** - Normalized RMS values of EMG signals of subjects with SCI (A, Left N=12; Right N=11) and control subjects (B, N=10) for left and right biceps femoris, during stance and swing phase, per condition, compared to the start baseline.
4.3.3 EMG - Gastrocnemius Medialis

No significant differences were found in the post-hoc analysis of the RMS values for the gastrocnemius medialis in subjects with SCI (see Figure 4.5A). In control subjects, significant differences were found in the left leg during stance and swing phase between the Steady and the Race condition ($p < 0.008$) (see Figure 4.5B). One control subject had to be removed from the analysis because of a loosened EMG electrode.

Figure 4.5: Gastrocnemius Medialis - Normalized RMS values of EMG signals of subjects with SCI (A, N=12) and control subjects (B, N=9) for left and right gastrocnemius medialis, during stance and swing phase, per condition, compared to the start baseline.
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4.3.4 EMG - Rectus Femoris

Post-hoc analysis of the RMS values for the rectus femoris in subjects with SCI showed significant differences in the left leg during stance phase between the Steady and the Speed ($p = 0.003$) and between the Steady and the Sprint condition ($p = 0.005$). In addition, significant differences were found during swing phase between the Steady and the Speed ($p = 0.002$), the Steady and the Sprint ($p = 0.006$) and between the Steady and the Race condition ($p = 0.005$) (see Figure 4.6A). In control subjects, post-hoc analysis revealed significant differences on the left side during swing phase between the Steady and the Speed ($p < 0.008$), the Steady and the Sprint ($p < 0.008$) and between the Steady and the Race condition ($p < 0.008$). On the right side, a significant difference was found during stance phase between the Steady and the Speed condition ($p < 0.008$) (see Figure 4.6B). One control subject had to be removed from the analysis because of a loosened EMG electrode.

4.3.5 EMG - Tibialis Anterior

Post-hoc analysis of the RMS values for the tibialis anterior in subjects with SCI showed significant differences in the left leg during stance phase between the Steady and the Speed ($p = 0.006$), the Steady and the Sprint ($p = 0.002$), the Steady and the Race ($p = 0.005$) and between the Speed and the Sprint condition ($p < 0.008$). A significant difference was found between the Sprint and the Race condition ($p = 0.023$) in the left leg. During the swing phase of the left leg, a significant difference was found between the Steady and the Sprint condition ($p = 0.003$) (see Figure 4.7A). In control subjects, post-hoc analysis revealed significant differences on the left side during stance phase between the Steady and the Sprint ($p = 0.009$) and between the Steady and the Race condition ($p = 0.007$). On the right side, significant differences were found during stance phase between the Steady and the Speed ($p = 0.005$), the Steady and the Sprint ($p = 0.009$) and between the Steady and the Race condition ($p = 0.009$) (see Figure 4.7B).
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Figure 4.7: Tibialis Anterior - Normalized (to start baseline) RMS values of EMG signals of subjects with SCI (A, N=12) and control subjects (B, N=10) for left and right tibialis anterior, during stance and swing phase, per condition, compared to the start baseline.
4.4 Discussion

Spinal cord injuries cause severe impairments of motor, sensory, and autonomic functions. Active participation during rehabilitation has shown to promote cortical plasticity through cortical map reorganization (Lynskey et al., 2008). Since the level of engagement influences active participation (Lequerica and Kortte, 2010), the aim of the study in this chapter was, to compare whether AF exercises with differing design characteristics can influence the level of physical engagement of patients. AF exercises were presented while subjects with and without SCI walked in a gait orthosis. I argued that an increase in engagement could be seen by an increase in physical activity, i.e., muscle activity and heart rate. Due to possible autonomic dysfunction in subjects with SCI (Bravo et al., 2004), heart rate measurements were interpreted with caution. Since muscle activation patterns have been shown to be different during walking in driven gait orthosis (Hidler and Wall, 2005) and SCI subjects exhibit different patterns of muscle activity than control subjects, comparing EMG between the two groups as well as with EMG during normal walking was omitted. Hence, I compared differences in muscle activity between exercises.

Out of all findings, the most prominent effect was found between AF exercises that differed in providing functionally meaningful reactions to motor performance. Thus, although not present in all muscles, our results show significant differences between interactive (i.e., Speed, Sprint, Race) and non-interactive (i.e., Steady) exercises. These were most prominent during the swing phase of the left rectus femoris for both subjects with and without SCI. Since the virtual speed was controlled by the hip joints during the swing phase, these significant differences are in great favor of the first hypothesis. Considering the additional differences in heart rate, I argue, that the first hypothesis, i.e., that AF exercises increase patient engagement only if they are interactive, can be confirmed. While previous studies were already able to show that the addition of AF exercises during walking in a gait orthosis increases muscles activity (Brütsch et al., 2010, Schuler et al., 2011), I, thus, further was able to verify that this increase only occurs, if the deployed AF exercise is interactive. In addition to increased engagement, the deployment of interactive AF exercises may additionally help to reduce the discomfort that today’s gait orthosis can induce (Borggraefe et al., 2010). Hence, research
4. Increasing Engagement using Augmented Feedback Exercises

has shown that interactive VR is capable of reducing pain (Wender et al., 2009). This, however, is subject to further studies.

Contrary to the first hypothesis, measurements did not support the second hypothesis, i.e., that AF exercises with explicit task goals and frequent performance feedback increase engagement compared to AF exercises without. Thus, although significant differences were found between the Speed and Sprint conditions in the right biceps femoris and the left tibialis anterior of subjects with SCI, I argue that these findings are rather arbitrary and, therefore, not adequate to sustain the second hypothesis. A lack of increased engagement could be attributed to multiple reasons. Thus, since research has shown that it is possible to estimate self-motion through optic flow in real (Frenz and Lappe, 2005) and virtual environments (Banton et al., 2005), the immediate feedback gained through virtual velocity changes may potentially have sufficed subjects during the Speed condition to rate their current level of performance thus reducing the need for further feedback mechanisms. The lack of explicit task goals in the Speed condition might additionally have encouraged a self-paced exploration of the VR environment (Ryan and Deci, 2000, Wood et al., 2004). Hence, while subjects were explicitly challenged to higher speeds in the Sprint and Race conditions, they might have produced similar ones in the Speed condition. Since research has shown, that pursuing self-defined goals can have a substantial effect on the level of effort, i.e., performance during a task (Bandura, 1991), the self-defined goal of exploring as much as possible of the VR environment may, therefore, have led to increased active participation.

Finally, in contradiction to the third hypothesis, i.e., that competitive AF exercises increase engagement compared to non-competitive exercises, there was only one significant difference between the Race and the Sprint condition in the left tibialis anterior of subjects with SCI. Contrary to the hypothesis, the muscle activity in the Race condition was, however, lower. The most intuitive explication for these findings can be seen in the nature of the exercises themselves. Compared to the Sprint condition, I denoted the Race condition as competitive. However, while subjects were competing against a virtual opponent in the Race condition, they were competing against themselves in the Sprint condition. Since, during therapy, subjects with SCI mainly compare themselves to their own motor improvements (Wressle et al., 2002), they might have preferred the
4.4 Discussion

Sprint to the Race condition. In addition, although adapted to the abilities of the subject, the Race condition might have created a competitive pressure (Ryan and Deci, 2000). Since it is known, that social comparison can have both beneficial and detrimental effects, subjects that were always ahead of the opponent may, hence, have displayed higher active participation, due to derived satisfaction of mastering the challenge, whereas subjects that were always near or behind the opponent, lost interest in the activity and reduced their effort to actively participate (Bandura, 1991). At last, similar to other studies (König et al., 2008a, Brütsch et al., 2010), subjects may have increased their activity once the opponent was near and decreased it again once he was behind them. This might have led to an overall lower activity.

One can argue, that the AF exercises might have been too similar to elicit different levels of engagement. However, apart from the fact that video game characteristics are rated differently (Westwood and Griffiths, 2010), studies have further shown user and gender specific preferences when it comes to video game playing (Lazzaro, 2008, Sherry, 2004). Hence, similar effects, i.e., preferences of the subjects, could also have caused these inseparable results.

4.4.1 Study Limitations

The six different conditions, being presented for four minutes each, together with the relaxation periods of three minutes, resulted in a total duration of roughly 40 minutes. This constitutes the maximum training duration that is usually suggested for training without having exhaustion effects (~30-45min). Thus, although longer durations of each condition would have been favored for analysis, four minutes was the maximum that was possible. However, the presentation time of four minutes in the present study was longer than in other, similar studies (Brütsch et al., 2010).

Conditions were presented in a pseudorandom order, i.e., while the Steady and Speed conditions were presented in a random order during the first half, the Sprint and Race conditions were randomly presented during the second half of the experiment. Although this could have caused an ordering effect, the current approach was favored to avoid an early exertion of the subjects, since the Sprint and Race conditions might have resulted in higher physical activity.
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The uneven distribution of significant differences between the left and right legs, could be attributed to the fact, that subjects were able to interact with the AF exercises using both their hip joints during swing phase. Therefore, subjects may have used their dominant leg more frequently or intensively. Unfortunately, this might have suppressed the finding of additional significances. I, however, chose this approach, to reduce the effort needed to interact with the VR environment and thus, decrease the probability of early exhaustion. This limitation should be addressed when creating AF exercises for everyday, clinical motor rehabilitation.

Despite having an identical virtual environment, exercises differed in the use of audiovisual feedback. Hence, compared to the Steady and Speed conditions, the Sprint condition used audiovisual cues to inform subjects about their average speeds and the Race condition presented positive and negative audiovisual cues, when subjects passed or were passed by the competitor, respectively. The addition of this additional feedback was part of the second hypothesis. Nevertheless, I cannot exclude, that the differing types of audiovisual feedback given in the Sprint and Race conditions could potentially have had an additional effect on results (Bradley and Lang, 2000, Timmermans et al., 2009). Since the effects of differing audiovisual feedback on active participation during motor rehabilitation have, however, so far not been addressed, I currently cannot determine the degree of influence. Hence, this should be subject to further studies.

Finally, the number of subjects was limited. Additional subjects, especially in the SCI group, may have led to a more homogeneous distribution of age and level of impairment. Nonetheless, I do not believe that the rather heterogeneous distribution of age in the SCI group, had an effect on the measured results. Hence, independent of age, SCI subjects show high motivation to improve their motor skills.

4.4.2 Graphics

According to Wood et al. (2004), realistic, high-quality graphics are deemed to be an important structural characteristic of modern video games. I postulated, that the visual stimuli of an AF exercise should be adapted to the cognitive capabilities of the patients.
4.5 Conclusion

(see section 1.5). Since subjects that participated in the current study did not show perceptual deficits, I chose to use high-quality graphics.

The VR environment was well received and subjects positively commented on the presented exercises. Especially subjects with SCI favored the approach of using AF exercises during therapy. Subjects mentioned additional objects that could be added to make it even more interesting, e.g., adding a lake or having a nice panoramic view as a reward of “climbing” the virtual hill. This feedback was highly appreciated and will be considered in future AF exercise development.

4.5 Conclusion

By varying the design characteristics for AF exercises set forth in the introduction (see section 1.5), I was capable of inducing different levels of physical engagement in subjects training with the developed AF exercises. Since I was not able to confirm all of the hypotheses, these findings highlight the importance of further investigations on the characteristics that AF exercises should inherit and, in accordance with Flores et al. (2008), illustrate the need of thoroughly analyzing AF exercises regarding their effectiveness for motor rehabilitation. In summary, however, the results of this study show, that functional feedback is highly important for the active participation of patients during robot-aided rehabilitation. Since high levels of effort are thought to be important for facilitating motor learning, functional feedback may, hence, positively affect the overall effectiveness of therapy (Kaelin-Lang et al., 2005, Reinkensmeyer et al., 2009).

4.6 Contribution

All methods presented in this chapter were developed during my thesis. Patient data was recorded by Mario Jacky at the “Medical Faculty, Balgrist University Hospital, University of Zurich, Zurich, Switzerland”.

Most previous studies only assessed the deployment effect of AF exercises, i.e., the effect that using or not using AF exercises has on engagement, or use AF exercises
4. Increasing Engagement using Augmented Feedback Exercises

as an amendment to robotic appliances, while design characteristics of the AF exercises were not evaluated. With the current study, I was able to show, that AF exercises have to be interactive, i.e., provide functionally meaningful reactions to the motor performance of the patient, to increase physical engagement during motor rehabilitation. No increases in engagement were, however, found for the design characteristics of explicit task goals, frequent performance feedback as well as competitive situations.

Considering the findings in chapter 2 and 3, I believe that these results further demonstrate, that it is important for future AF exercises to take user preferences and expectations into consideration (Sherry, 2004, Lazzaro, 2008, Westwood and Griffiths, 2010). Hence, user and gender specific preferences or, for instance, the possibility of self-definable in contrast to predefined task goals, may all have a subjective influence on the resulting level of active participation during motor rehabilitation.
5

Conclusion & Outlook

5.1 Conclusion

Up till now, there are no explicit guidelines of how augmented feedback (AF) exercises have to be designed to assure optimal motivation and, hence, increased engagement (Yim and Graham, 2007, Flores et al., 2008). Most studies in research only assess the deployment effect of AF exercises, i.e., the effect that using or not using AF exercises has on engagement, or use AF exercises as an amendment to robotic appliances. Although some studies did propose characteristics that should encourage motivation (Lövquist and Dreifaldt, 2006, Pareto et al., 2008), their actual effectiveness to increase engagement during therapy has so far not been verified.

Hence, the overall aim of this thesis was to assess the effect that AF exercises have on motivation and investigate what characteristics they need to include, to increase active participation, i.e, patient engagement, during robot-aided motor rehabilitation.

The current investigations particularly focused on the flow-derived design characteristics set forth in section 1.5. Hence, I postulated that AF exercises should (a) be able to adapt their level of difficulty to the capabilities of the patient, (b) provide explicit task goals, making therapy benefits apparent, (c) provide frequent performance feedback, (d) allow adjustable visual stimuli, (e) show functionally meaningful reactions to the motor performance of the patient and (f) allow the possibility to accomplish competitive training, in order to increase engagement.
5. Conclusion & Outlook

Since I assumed that adapting the level of difficulty to the capabilities of the patient is a design characteristic of high importance, I specifically addressed this characteristic in more detail.

Although measuring the effects of different visual stimuli on patient engagement is equally important, it is an extensive task that has to be addressed in more detail and, thus, should be the subject to further studies.

5.1.1 Augmented Feedback Exercises & Patient Engagement

Patient engagement can be defined as a construct that is driven by motivation and executed through active, effortful participation during therapy (Lequerica and Kortte, 2010). It is a multidimensional construct that contains cognitive, behavioral, i.e., physical and emotional components (Kahn, 1990, Lequerica and Kortte, 2010). Since the emotional component of engagement mostly depends on the psychological state of the patient, the current studies mainly focused on the aspects of cognitive and physical engagement of deploying AF exercises during robot-aided motor rehabilitation.

With the first study, I was able to show that AF exercises increase the active participation, i.e., engagement of patients during therapy. This was reported by patients, but also perceived by their therapists. Although the graphics of the AF exercises played a secondary role, the feedback from patients did show the broad range of demands patients have towards the visual stimuli of the virtual environment. Another finding was, that it is challenging for therapists to manually adjust the task difficulty of an AF exercise to the capabilities of the patient. Since this could affect cognitive engagement it was, thus, addressed in more detail with the second study.

The second study mostly focused on the cognitive aspects of AF exercises. Cognitive engagement may be influenced by a persons cognitive abilities in relation to task specific characteristics, e.g., the level of difficulty of a task (Macey and Schneider, 2008). Results showed that increasing task difficulty indeed was followed by a perceived increase of mental demand resulting in decreased performance. In addition, subjects reported increasing levels of frustration. Hence, the level of difficulty may not only affect cognitive but also lead to emotional disengagement (Lequerica and Kortte, 2010).
Therefore I conclude that it is crucial to adapt the level of difficulty of an AF exercise to the capabilities of the patient. This will increase engagement and thus active participation. While the current study was focused on upper-extremity rehabilitation, studies about task difficulty have also been performed for lower-extremity rehabilitation (König et al., 2011a).

Finally, the last study focused on influencing physical engagement. Results indicated that AF exercises have to provide functionally meaningful reactions to the motor performances of the patients, in order to increase active participation. Providing explicit task goals, frequent performance feedback or competitive situations did not have any significant effects. This absence of increased active participation again might be attributed to the emotional component of engagement. Hence, patients showed differing preferences and expectations towards the AF exercises (Sherry, 2004, Westwood and Griffiths, 2010). While for example some subjects favored competition, others were too stressed and might have preferred a non-competitive, self-paced exploration of the virtual environment (Ryan and Deci, 2000, Wood et al., 2004).

In summary, these findings illustrate that AF exercises are indeed capable of affecting and partially increasing cognitive, physical and potentially emotional engagement during motor rehabilitation. They, however, further emphasize the need of developing guidelines that describe the effects that different design characteristics have on engagement during rehabilitation.

### 5.1.2 Design Characteristics for Augmented Feedback Exercises

In the current studies, not all of the derived design characteristics induced significant changes in engagement. Effects could only be found for augmented feedback exercises that adapt the level of difficulty to the capabilities of the patient and exercises that show functionally meaningful reactions to the motor performance of the patient. These should therefore be fully respected during future developments, and if possible supplemented to existing AF exercises. Nevertheless, I still suggest that all of the flow-derived design characteristics should be considered when developing AF exercises for robot-aided motor rehabilitation. Hence, while adaptive difficulty and interactivity
5. Conclusion & Outlook

mostly aim at cognitive and physical engagement, the others could have a potential effect on the emotional component.

Due to the residual amount of possible design characteristics, further investigations are, however, still indispensable. Different aspects will be highlighted in the Outlook (see section 5.2).

5.1.3 Difficulty Adaptation using Fitts’ Law

Independent of measuring the effects of AF exercises on patient engagement, I additionally was able to successfully demonstrate that Fitts’ Law can be used as mechanism to adjust the level of difficulty of a task to the capabilities of the patient. Due to its simplicity, the mechanism can be easily incorporated into a large number of existing and novel upper-extremity, two-dimensional, pointing AF exercises. By varying the distance and size of start and target locations, Fitts’ Law additionally allows an unlimited number of variations and, thus, a high diversification of visual stimuli in AF exercises.

5.2 Outlook

Due to the increasing number of patients, the pressure of the healthcare system to reduce medical costs and limited therapeutic resources, clinicians may in the future want patients to practice their movements at home or with reduced supervision during their stay in the clinic. The need for AF exercises that are capable of autonomously keeping a patient engaged and motivated while respecting a patients’ capabilities will hence rise. Remedy may come from AF exercises that are not only capable of adapting their level of difficulty to the physical capabilities, but also their visual stimuli to the cognitive capabilities of the patient.

Current AF exercises mainly make use of sensory input from the rehabilitation device, various exercise metrics (e.g. Score, Events of Success or Failure) and sometimes psycho-physiological measurements, to measure motivation and physical capabilities and adapt the level of difficulty accordingly (see Figure 5.1). In my view, future exer-
Cises should also assess the preferences and cognitive capabilities of patients, adapting the audiovisual stimuli (e.g., object shapes, colors or sounds) accordingly.

Figure 5.1: Interactions between Patient and Augmented Feedback Exercise - The goal and type of an AF exercise affect the perceived benefit and preferences, i.e., expectations of a patient. This eventually has an influence on the motivation of the patient to actively participate using the corresponding exercise. Motivation, however, is also influenced by the progress in the AF exercise, which is used to evaluate the perceived outcome. A patient’s motivation, in combination with his/her capabilities finally influences the overall physical activity, i.e., engagement in the rehabilitation device to control the AF exercise. Novel AF exercises should not only use psycho-physiological assessments, sensory input from the rehabilitation device and exercise metrics to measure the performance and adapt the level of difficulty to the patient’s capabilities, but also adjust the audiovisual stimuli to the expectations of the patient (e.g., through methods like A/B Testing).

Cognitive capabilities could, for instance, be measured by incorporating well-known assessments directly into the AF exercise (e.g., the star cancellation test; Halligan et al. 1990). Using this example, one could online assess unilateral neglect and, thus, adapt the audiovisual stimuli accordingly. Additional assessments could be directed towards attention, memory, visuomotor and visuospatial cognitive functions (Rizzo and
5. Conclusion & Outlook

Buckwalter, 1997).

Regarding preferences, AF exercises could make use of methods like A/B Testing in order to assess and adapt to the most effective audiovisual stimuli (Kohavi et al., 2007), i.e., interchange different stimuli and use exercise metrics to determine the optimal configuration for a particular patient.

Finally, I suggest further investigations into optimizing exercise metrics. While many approaches use psycho-physiological measurements to adapt AF exercises to the patient, studies have shown that exercise metrics are more reliable in estimating a patient’s level of engagement (Novak et al., 2011). Since psycho-physiological measurements require additional, complex analysis, produce additional costs, increase the overall complexity of the system and are difficult to apply during everyday, clinical therapy, finding ways to correctly adapt the exercise based on its own metrics would be of high value.
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5.1 **Interactions between Patient and Augmented Feedback Exercise** -
The goal and type of an AF exercise affect the perceived benefit and preferences, i.e., expectations of a patient. This eventually has an influence on the motivation of the patient to actively participate using the corresponding exercise. Motivation, however, is also influenced by the progress in the AF exercise, which is used to evaluate the perceived outcome. A patients motivation, in combination with his/her capabilities finally influences the overall physical activity, i.e., engagement in the rehabilitation device to control the AF exercise. Novel AF exercises should not only use psycho-physiological assessments, sensory input from the rehabilitation device and exercise metrics to measure the performance and adapt the level of difficulty to the patients capabilities, but also adjust the audiovisual stimuli to the expectations of the patient (e.g., through methods like A/B Testing). ........................................... 81
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<tr>
<td>ACA</td>
<td>Anterior Cerebral Artery</td>
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<td>AF</td>
<td>Augmented Feedback</td>
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<td>AIS</td>
<td>American Spinal Injury Association Impairment Scale</td>
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<td>ANOVA</td>
<td>Analysis of Variance</td>
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<td>BDI</td>
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<td>BWS</td>
<td>Body Weight Support</td>
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<td>CP</td>
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<td>Driven Gait Orthosis</td>
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<td>EEG</td>
<td>Electroencephalography</td>
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<td>EHI</td>
<td>Edinburgh Handedness Inventory</td>
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<td>Experience Sampling Method</td>
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<td>Good Clinical Practice</td>
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<td>Guidance Force</td>
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<td>HPR</td>
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<td>International Conference on Harmonisation of Technical Requirements for Registration of Pharmaceuticals for Human Use</td>
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<td>ID</td>
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