A comprehensive assessment of the musculoskeletal system: The CAMS-Knee data set

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A B S T R A C T

Combined knowledge of the functional kinematics and kinetics of the human body is critical for understanding a wide range of biomechanical processes including musculoskeletal adaptation, injury mechanics, and orthopaedic treatment outcome, but also for validation of musculoskeletal models. Until now, however, no datasets that include internal loading conditions (kinetics), synchronized with advanced kinematic analyses in multiple subjects have been available. Our goal was to provide such datasets and thereby foster a new understanding of how in vivo knee joint movement and contact forces are interlinked – and thereby impact biomechanical interpretation of any new knee replacement design. In this collaborative study, we have created unique kinematic and kinetic datasets of the lower limb musculoskeletal system for worldwide dissemination by assessing a unique cohort of 6 subjects with instrumented knee implants (Charité – Universitätsmedizin Berlin) synchronized with a moving fluoroscope (ETH Zürich) and other measurement techniques (including whole body kinematics, ground reaction forces, video data, and electromyography data) for multiple complete cycles of 5 activities of daily living. Maximal tibio-femoral joint contact forces during walking (mean peak 2.74 BW), sit-to-stand (2.73 BW), stand-to-sit (2.57 BW), squats (2.64 BW), stair descent (3.38 BW), and ramp descent (3.39 BW) were observed. Internal rotation of the tibia ranged from 3° to 9.3° internal. The greatest range of antero-posterior translation was measured during stair descent (medial 9.3 ± 1.0 mm, lateral 7.5 ± 1.6 mm), and the lowest during stand-to-sit (medial 4.5 ± 1.1 mm, lateral 3.7 ± 1.4 mm). The complete and comprehensive datasets will soon be made available online for public use in biomechanical and orthopaedic research and development.

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1. Introduction

Accurate knowledge of the internal loading conditions in the human musculoskeletal system forms the basis for understanding a wide range of biomechanical processes including musculoskeletal adaptation (Szwedowski et al., 2012; Thompson et al., 2017), orthopaedic treatment outcome (Niki et al., 2013), wear and failure mechanisms (Argenson and Parratte, 2006), overloading and injury mechanics (Boeth et al., 2013), as well as for optimising implant designs, and validating musculoskeletal models (Schellenberg et al., 2017). However, many aspects of modelling and understanding biomechanical interactions in the human musculoskeletal system, are limited by the lack of availability of complete and synchronous kinematic and kinetic datasets. In their “Grand-Challenge”, Fregly and co-workers annually released musculoskeletal datasets based on data collected from a single subject implanted with a force-measuring knee replacement (Fregly et al., 2012; Kinney et al., 2013). The distribution of these datasets signified a landmark in the ability of the entire musculoskeletal modelling community worldwide to use this

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data and validate their own lower limb models. However, the limited number of subjects and datasets available has restricted any population or activity based modelling. Furthermore, kinematics of the limbs were primarily extracted from optical motion capture (only walking on a treadmill was measured fluoroscopically), which is subject to soft tissue artefact (Taylor et al., 2005), and the accuracy of the kinematic assessment therefore clearly limited the ability to understand the role of tibio-femoral motion on the internal joint contact forces.

Video-fluoroscopy allows the accurate reconstruction of objects with a known geometry in 3D space, and has thus become a well-accepted imaging technique to acquire kinematic information of artificial joints during simple functional movement tasks such as squatting or rising from a chair. However, measurements during functional activities of daily living such as walking and stair descent have remained extremely limited: the heavy physical structure of the imaging technology has generally restricted the development of mobile devices. As such, only a handful of mobile units exist that enable the tracking of moving joints. The Laboratory for Movement Biomechanics, ETH Zürich, has developed a single plane moving fluoroscope that is capable of tracking human joints throughout complete cycles of activities of daily living (Zihlmann et al., 2006b) (Fig. 1), thus allowing the accurate reconstruction of the 3D kinematics of e.g. the knee joint (Banks and Hodge, 1996; Hoff et al., 1998; Zhu and Li, 2012) without inaccuracies associated with soft tissue artefact.

The development of telemetric implants at the Charité – Universitätsmedizin Berlin, Germany, has allowed improved understanding of the internal loading conditions that occur in subjects with artificial joints. Using strain gauges fixed within the shaft of the tibial component, this technology allows the tibio-femoral forces and moments that act within the implant to be captured during dynamic activities in the knee joints in vivo (Heinlein et al., 2007). Comprehensive information about the loading of orthopaedic implants is already provided in the Orthoload database (www.orthoload.com). However, until now, this data has mainly been limited to joint kinematics. In this respect, expansive data for multiple subjects that includes accurate information of both joint kinetics and kinematics remains elusive.

In a unique collaborative effort, the “Comprehensive Assessment of the Musculoskeletal System” (CAMS-Knee) project aimed to unite these technologies and capture synchronous datasets of kinematics and kinetics of the human knee. With the goal to make these datasets widely available, it was our aim to support the field of musculoskeletal biomechanics and provide researchers and industry a reliable and highly accurate resource for model validation and research into the movement and loading of the human knee, particularly in subjects with total knee replacements.

Table 1
Description of the activities performed.

<table>
<thead>
<tr>
<th>Activity</th>
<th>Description</th>
</tr>
</thead>
<tbody>
<tr>
<td>Level walking</td>
<td>Walking straight ahead over 5 force plates embedded in the floor</td>
</tr>
<tr>
<td>Downhill walking</td>
<td>Walking down a walkway with a 10° inclined slope (18%), which included two force plates</td>
</tr>
<tr>
<td>Stair descent</td>
<td>Walking down an instrumented stair with three steps, each 18cm in height</td>
</tr>
<tr>
<td>Sitting and rising from a chair</td>
<td>The two tasks stand-to-sit and sit-to-stand were measured as a single sequence. The subject started in a sitting position, rose to an upright standing position and sat down again</td>
</tr>
<tr>
<td>Squat</td>
<td>Standing with stationary feet, approximately shoulder width apart with hands stretched forwards Knee joint flexion as far as individually possible before returning to the standing position</td>
</tr>
</tbody>
</table>

2. Materials and methods

2.1. Subjects

Six subjects (5 m, 1 f, aged 68 ± 5 years, mass 88 ± 12 kg, height 173 ± 4 cm) each with an instrumented TKA (Heinlein et al., 2007) were measured approximately 16 years post-operatively while performing multiple repetitions of different activities of daily living Table 1. All testing of subjects involved within this project were performed in accordance with the Declaration of Helsinki. The study was approved by the local ethics committees of the Charité (EA/065/06) and ETH Zürich (EK 2013-N-90) and all subjects provided written informed consent prior to participation.

2.2. Telemetry

Each subject possessed a cemented INNEX knee implant (Zimmer, Switzerland; type FIXUC), in which the tibial component was modified and instrumented with 6 semi-conductor strain gauges and a 9-channel telemetry transmitter (90-100Hz), allowing six-component load measurements of the 3 contact forces and 3 joint moments acting on the tibial component to be recorded with a mean measurement error below 2% (Heinlein et al., 2007). All signals were sensed and transmitted using a custom-made, inductively powered telemetry circuit (Graichen et al., 2007). During measurements, the subjects wore an external coil and antenna around the shank, which were connected to a custom-made receiver and amplifier. The signals were recorded together with the patient video on a digital video tape and prepared for post-processing evaluation. One of the audio tracks recorded the demodulated pulse trains of the telemetry signals and the other the synchronization signal. Finally, all forces and moments were determined and presented in the tibial coordinate system (detailed in Fig. 1).

2.3. Fluoroscopy

To overcome the limitations of marker-based kinematic measurements, which are affected by soft tissue artefacts (Taylor et al., 2005; Taylor et al., 2010), tracking fluoroscopic techniques for dynamically imaging internal skeletal structures and metallic implant components have been developed (Foresti, 2009; Zihlmann et al., 2006a; Zihlmann et al., 2006b). The video-fluoroscope c-arm unit was mounted on an automated trolley (maximum acceleration horizontal 9 ms⁻², maximum velocity horizontal 5 ms⁻¹) that allows dynamic tracking of the joint in question. The additional ability of the C-arm to track the joint at up to 1.33 ms⁻¹ vertically (maximum acceleration 4 ms⁻²), thus

Fig. 1. Coordinate system of the instrumented tibial tray. Figure adapted from (Kutzner et al., 2010) and reprinted with permission.
enabled multiple cycles of free level and downhill walking as well as stair descent (List et al., 2017b) to be captured.

The video fluoroscopic image capture was performed using a modified BV Pulsera videofluoroscopy system (Philips Medical Systems, Switzerland) with a field of view of 30.5 cm, pulsed image acquisition rate of 25 Hz, 8 ms radiation time, 1 ms shutter time of the CCD-sensor and an image resolution of 1000 × 1000 pixels with a grayscale resolution of 12 bits (Foresti, 2009; List et al., 2012a; List et al., 2012b; Zihlmann et al., 2006a; Zihlmann et al., 2006b).

Image distortion of the videofluoroscopic images was eliminated by a local correction algorithm (Foresti, 2009; List et al., 2012a; List et al., 2012b) using a reference grid containing approximately 1300 beads. Since the relative position of the beads was known, the projection of the reference grid was restored by means of a polynomial approximation. The projection parameters of the videofluoroscopic system (focal distance and location of the principle point in the image plane) were determined using a least-squares optimization, which was based on five images of a calibration tube (300 mm long with two Plexiglas® plates). At well defined positions (accuracy: ±0.03 mm), each plate was filled with either 12 or 13 metal pellets, providing a total of 25 correspondence points.

Once the projection parameters of the fluoroscope were determined, its orientation and location relative to the video-photogrammetric system were determined (Foresti, 2009; List et al., 2017b). Here, the grid used for the image distortion correction was also equipped with six reflective markers screwed at predefined positions. The grid was rotated and displaced into multiple poses, with radiographs of the grid’s beads and simultaneous assessment of the marker positions allowing the relationship between their local coordinate systems to be determined. In order to determine the projection matrix, a least squares optimization was used to find the orientation and position of the fluoroscopy system relative to the origin of the video-photogrammetric setup. Optical markers were additionally fixed to the C-arm of the moving fluoroscope to allow the position of the moving fluoroscope to be continually determined and referenced to the global laboratory coordinate system.

2D/3D registration of the 2D fluoroscopic images was performed by fitting CAD models of the implant components. The registration algorithm was based on the approach developed by Burckhardt et al. (2005), in which the pose of the 3D implant CAD models was determined through fitting a synthetic image of the CAD model to the fluoroscopic image by minimizing the difference in gradient magnitudes as well as pixel grey values within the region of interest defined by a slightly enlarged outline contour, to create the optimal matching scenario for each time point. Registration errors, assessed for a similar TKA, were <1 degree for all rotations, <1 mm for in-plane and <3 mm for out-of-plane translations.

Fig. 2. Positions of the reflective markers. The naming convention relates to the description of the placement in Table 2.
**Table 2**

Description of the skin marker placement.

<table>
<thead>
<tr>
<th>Segment</th>
<th>Marker position</th>
<th>Marker name</th>
</tr>
</thead>
<tbody>
<tr>
<td>Trunk/Arm</td>
<td>Seventh cervical vertebra</td>
<td>C7</td>
</tr>
<tr>
<td></td>
<td>Highest point of the acromion</td>
<td>RTSH/LTSH</td>
</tr>
<tr>
<td></td>
<td>Epicondylo radial</td>
<td>RTEL/LETLE</td>
</tr>
<tr>
<td></td>
<td>Styloid process of radius</td>
<td>RTKS/LTRS</td>
</tr>
<tr>
<td></td>
<td>Ulnar styloid process</td>
<td>RTUS/LTUS</td>
</tr>
<tr>
<td>Pelvis</td>
<td>Sacrum</td>
<td>SACR</td>
</tr>
<tr>
<td></td>
<td>Posterior superior iliac spine</td>
<td>RTPS/LTPS</td>
</tr>
<tr>
<td></td>
<td>Crista iliaca, dorsal</td>
<td>RTPE/LTPE</td>
</tr>
<tr>
<td></td>
<td>Mid superior iliac spine</td>
<td>RTMS/LTMS</td>
</tr>
<tr>
<td></td>
<td>Anterior superior iliac spine</td>
<td>RTAS/LTAS</td>
</tr>
<tr>
<td>Thigh</td>
<td>Lateral thigh on 50% thigh length</td>
<td>RTTL/LTTL</td>
</tr>
<tr>
<td></td>
<td>Lateral thigh on 20% thigh length</td>
<td>RTTL/LTTL</td>
</tr>
<tr>
<td></td>
<td>Lateral epicondyle</td>
<td>RTLE/LTLE</td>
</tr>
<tr>
<td></td>
<td>Medial epicondyle</td>
<td>RTME/LTME</td>
</tr>
<tr>
<td></td>
<td>Front thigh, one hand above knee</td>
<td>RTFR/LTFR</td>
</tr>
<tr>
<td></td>
<td>Ventral thigh on 50% of the length</td>
<td>RTAT/LTAT</td>
</tr>
<tr>
<td></td>
<td>Upper 1/3 of the dorsal thigh</td>
<td>RTPP/LTPP</td>
</tr>
<tr>
<td></td>
<td>Lower 1/3 of the dorsal thigh</td>
<td>RTPD/LTPD</td>
</tr>
<tr>
<td>Shank</td>
<td>Head of fibula</td>
<td>RTHF/LTHF</td>
</tr>
<tr>
<td></td>
<td>Tibial tuberosity</td>
<td>RTTT/LTTT</td>
</tr>
<tr>
<td></td>
<td>Mid tibia on 50% shank length</td>
<td>RTMT/LTMT</td>
</tr>
<tr>
<td></td>
<td>Lower 1/3 of the ventral shank</td>
<td>RTDT/LDTT</td>
</tr>
<tr>
<td></td>
<td>Lateral fibula on 30% shank length</td>
<td>RTLF/LTLF</td>
</tr>
<tr>
<td></td>
<td>Upper 1/3 of the lateral shank</td>
<td>RTPS/LTPS</td>
</tr>
<tr>
<td></td>
<td>Lower 1/3 of the dorsal shank</td>
<td>RTPD/LTPD</td>
</tr>
<tr>
<td></td>
<td>Lateral malleolus</td>
<td>RTLM/LTLM</td>
</tr>
<tr>
<td></td>
<td>Medial malleolus</td>
<td>RTMM/LTMM</td>
</tr>
<tr>
<td>Rear foot</td>
<td>Calcaneus lateral below lateral malleolus</td>
<td>RTLC/LTLC</td>
</tr>
<tr>
<td></td>
<td>Calcaneus posterior inferior</td>
<td>RTTH/LTTH</td>
</tr>
<tr>
<td></td>
<td>Calcaneus anterior superior</td>
<td>RTBH/LTHH</td>
</tr>
<tr>
<td></td>
<td>Calcaneus lateral below medial malleolus</td>
<td>RTMC/LTMC</td>
</tr>
<tr>
<td>Forefoot</td>
<td>Base of fifth metatarsal</td>
<td>RTVB/LTVB</td>
</tr>
<tr>
<td></td>
<td>Head of fifth metatarsal</td>
<td>RTVM/LTVM</td>
</tr>
<tr>
<td></td>
<td>Head of second metatarsal</td>
<td>RTTO/LTTO</td>
</tr>
<tr>
<td></td>
<td>Head of first metatarsal</td>
<td>RTFP/LTFP</td>
</tr>
<tr>
<td></td>
<td>Base of first metatarsal</td>
<td>RTFB/LTFB</td>
</tr>
</tbody>
</table>

(Foresti, 2009). The output is the 3D pose of the tibial and femoral components relative to the lab or image intensifier coordinate systems according to Grood and Suntay (1983).

**2.4. Whole body kinematics**

To analyse the full body kinematics, a 3D motion capture system (Vicon, OMG, UK) consisting of twenty-six MX40 and T160 motion-capture cameras recorded the motion of 75 skin markers attached to the skin at 100 Hz. The markers were attached mainly to the lower extremities (Fig. 2; Table 2), and specifically encompassed all marker positions required for the fB marker set (List et al., 2013), the OSSCA bone landmark and cluster marker sets (Ehrig et al., 2011; Ehrig et al., 2006, 2007; Heller et al., 2011; Kratzenstein et al., 2012; Taylor et al., 2010), and the lower-limb Plug-in-Gait marker locations (Vicon Peak®, Oxford, UK) (see Table 3).

**2.5. Ground reaction forces**

Six force plates (B1 and B2, type 9281B, 400 × 600 mm, B3 and B4, type 9285, 400 × 600 mm, B5, type 9281C, 400 × 600 mm and A1, type 9287B, 600 × 900 mm, 2000 Hz; Kistler, Switzerland) aligned with the walkway, were used to measure the ground reaction forces (GRFs). These force plates were decoupled from the surrounding floor through their installation on an isolated concrete foundation (mounted directly on the ground floor below) to eliminate signal noise caused by ground vibration due to movement of the fluoroscope. The staircase and the ramp used for downhill walking were equipped with two mobile force plates (C1 and C2, type 9286AA, 400 × 600 mm, 2000 Hz; Kistler, Winterthur, Switzerland). To obtain the exact location of the origin, as well as the orientation of the mobile force plates, the position of calibration markers plugged into the force plates were captured.

All force plates were additionally calibrated to improve the estimation of the centre of pressure (CoP) with an in-situ point of force application calibration method (List et al., 2017a). As a result of the procedure, the mean error of the determined CoP was thereby reduced from 0.8 to 19.8 mm before correction to within a range of 0.04–2.2 mm.

**2.6. EMG**

The muscular activities and their coordinated responses were detected using a 16-channel wireless EMG system (Trigno, Delsys, USA), which was checked prior to subject measurements to ensure no interference from the implant telemetry data transfer. The EMG dual surface electrodes are placed on the preselected muscles to detect the myoelectric signals throughout the motion tasks. The recorded data was telemetrically sent to the workstation and synchronized with the kinematic measurements.

The electrodes were attached to the skin at eight predominant muscle sites on each lower limb (Fig. 3). At the beginning of the test session, the EMG signals during maximal voluntary contractions of each muscle were recorded. For this purpose, the following four motion tasks were performed for both legs:

**2.6.1. Triceps surae**

One legged standing together with lifting the heel to stand on tiptoe.

**2.6.2. Quadriceps**

With the subject sitting on a bench, with their lower legs hanging down, extension of the knee joint was performed against a load by means of a strap around the lower leg, just above the ankle.

**Table 3**

EMG electrode placement for both the left (channels 1–9) and right (channels 10–16) limbs.

<table>
<thead>
<tr>
<th>Channel</th>
<th>Muscle</th>
<th>Description of EMG electrode placement</th>
</tr>
</thead>
<tbody>
<tr>
<td>1/9</td>
<td>RectFem</td>
<td>50% from the anterior spina iliaca superior to the superior aspect of the patella (Perotto, 2011)</td>
</tr>
<tr>
<td>2/10</td>
<td>VastusMed</td>
<td>4 fingers superior of the superior part of the patella (Perotto, 2011)</td>
</tr>
<tr>
<td>3/11</td>
<td>VastusLat</td>
<td>One hand-width superior of the superior aspect of the patella (Perotto, 2011)</td>
</tr>
<tr>
<td>4/12</td>
<td>TibAnt</td>
<td>1/3 from the tip of the fibula to the tip of the medial malleolus</td>
</tr>
<tr>
<td>5/13</td>
<td>HamMed</td>
<td>50% between the medial epicondyle of the femur and the ischial tuberosity (Semitendinosus) (Perotto, 2011)</td>
</tr>
<tr>
<td>6/14</td>
<td>HamLat</td>
<td>50% between the fibula head and the ischial tuberosity (Biceps femoris long head) (Perotto, 2011)</td>
</tr>
<tr>
<td>7/15</td>
<td>GastroMed</td>
<td>One hand-width below the hollow of the knee on the medial muscle belly (Perotto, 2011)</td>
</tr>
<tr>
<td>8/16</td>
<td>GastroLat</td>
<td>One hand-width below the hollow of the knee on the lateral muscle belly (Perotto, 2011)</td>
</tr>
</tbody>
</table>
2.6.3. Hamstrings
With the subject sitting on a bench, with their lower legs hanging down, flexion of the knee joint was performed against a load by means of a strap around the lower leg, just above the ankle.

2.6.4. Tibialis anterior
With the subject standing, dorsiflexion of the ankle was performed against a manual resistant force provided by the investigator.

2.7. Video
Videos of each measurement were recorded using a digital camera (Panasonic NV-GS400) together with a digital video recorder (GV-D1000) for event documentation. The synchronization to the audio track, on which the telemetry data was stored, was performed by an LED light (delay ≤ 1 video frame).

2.8. Synchronisation
All measurement systems recorded simultaneously and were temporally synchronized. While the GRFs and EMG data were read directly into Vicon Nexus, fluoroscopic images were synchronized using a TTL trigger signal input into Vicon to temporally register each frame. In addition, Vicon was synchronized with the internal force measurement telemetry by sending a TTL trigger signal to the telemetry system.

2.9. Activities

2.9.1. Calibration tasks
Before the fluoroscopic measurements were performed, each subject performed basic motion tasks according to List et al. (2013), to allow functional determination of the joint centres at the hips, as well as axes of rotation for the knee and ankle joints.

Prior to measurement with the moving videofluoroscope, practice trials without imaging were performed until the subjects felt comfortable with the measurement systems and protocols. Free level gait, downhill walking, stair descent, sit-to-stand and stand-to-sit, as well as squatting activities, were then performed while all measurement systems were active Table 1. For each motion task, a minimum of five valid trials were acquired. For a valid trial, the knee had to be in the field of view of the image intensifier during the stance as well as the swing phase, and the force plates had to be hit correctly.

2.9.2. Level walking
Level walking included walking straight ahead over 5 force plates embedded in the floor (Fig. 3).

2.9.3. Downhill walking
A ramp, consisting of a walkway with a 10° inclined slope (18%) and two included embedded force plates (registered to the global coordinate system), was developed to perform downhill walking. Each subject started walking and the fluoroscope tracked to measure the final complete gait cycle of the downhill walking.

2.9.4. Stair descent
Walking downstairs consisted of descending a staircase of three steps, each 18 cm in height. Force plates were mounted within the stairs, each registered to the global coordinate system.

2.9.5. Sitting and rising from a chair
Sit-to-stand and stand-to-sit were both measured in one motion sequence. The subject started in a sitting position, rose to an upright standing position and sat down again. Subjects were seated at an angle to the fluoroscope to avoid interference from the second knee.

2.9.6. Squat
For the squat activity, subjects stood with stationary feet, approximately shoulder width apart, and hands stretched forwards. Knee joint flexion was then performed as far as possible before returning to the standing position.

2.10. Processing of the data
Tibio-femoral A-P translations and proximo-distal (p-d) distances were described using the distance-weighted means of the 10 nearest points on each of the medial and lateral femoral condyles (surface element edge length approx. 2.8 mm) relative to the tibial baseplate, which were re-calculated for each time point of the recorded kinematics to account for relative motion and rotation of the implant components. As a result, anterior translation of the medial contact point would denote internal tibial rotation.

For all gait activities, all kinematic and kinetic parameters were temporally normalized to a complete gait cycle. The gait cycle was defined from heel-strike to heel-strike. Heel-strikes and toe-offs were defined using a GRF threshold of 25 N. Mean and standard deviation of the parameters were extracted from at least five valid cycles of each activity and presented as a function of time normalized the activity cycle.

3. Results

In general, all subjects were able to successfully undertake all activities. Considerable variations in both kinematic and kinetic parameters were observed between subjects, but also between trials in individuals. Exemplary results of the kinematic and kinetic parameters obtained by moving video-fluoroscopy and the instrumented knee implant during different activities are presented for one subject (Fig. 4). However, it is our intention that complete datasets for all subjects, activities, trials and measurement modalities will be made freely available for non-commercial usage. Consequently, the following results are all presented as mean values over all subjects in order to provide a greater overview of the population kinematics and kinetics:

Knee flexion angles during level walking, downhill walking and stair descent exhibited a biphasic pattern for all subjects. The mean knee flexion ROM across all subjects for level walking was 56.0 ± 6.4°, 65.2 ± 3.0° for downhill walking, and 87.1 ± 4.4° for stair descent. The sit-to-stand and stand-to-sit activities resulted in a knee flexion to a mean of 76.0 ± 6.8°, and extended to a mean of −3.9 ± 11.2°. In a similar manner, the knee flexion reached a mean of 73.1 ± 9.4° during the squat activity.

The mean ad-abduction of the knee remained nearly constant throughout all activities and did not exceed 2°. External rotation of the tibia was rarely observed. Internal rotation of the tibia increased with increasing knee flexion for almost all activities, resulting in a relatively large RoM, ranging from 3° external (occurring during gait) to 9.3° internal (during squats) rotation. A-P translation of the nearest medial and lateral articular points (described above) demonstrated different patterns during different activities, but with similar RoMs. The greatest RoM was measured during stair descent (medial 9.3 ± 1.0 mm, lateral 7.5 ± 1.6 mm), while the lowest RoM was observed during the stand-to-sit activity (medial 4.5 ± 1.1 mm, lateral 3.7 ± 1.4 mm). The minimum p-d distance between the tibial and femoral components remained relatively constant, but some variations were visible during the walking task in this subject. The mean range of p-d distance was 1.4 ± 0.3 mm, 1.7 ± 0.6 mm, 2.0 ± 0.5 mm, 1.1 ± 0.2 mm, 1.3 ± 0.3 mm, 1.5 ± 0.5 mm for
level walking, ramp walking, stair descent, stand-to-sit, sit-to-stand, and squat respectively.

The tibio-femoral joint contact forces reached a mean peak of 2.74 BW during walking (highest single peak value of 3.73 BW was found in the database: Fig. 4), 2.73 BW during sit-to-stand, 2.57 BW during stand-to-sit, and 2.64 BW during the squat exercises. However, considerably higher forces of approximately 3.38 BW and 3.39 BW were observed during stair descent and ramp descent respectively. In general, the compressive forces followed different patterns during different activities; however, these patterns had nearly consistent shapes between subjects. As can be seen from the maximum and minimum values across all subjects and all trials (shown as "\" and "\" respectively for walking only; Fig. 4), considerable variability was observable between subjects. Of note was that the highest joint contact forces did not necessarily relate to the joint kinematics.

4. Discussion

Accurate knowledge of the relationship between joint kinematics and kinetics in vivo forms the foundation of understanding and improving many clinical and rehabilitation treatments in the fields of orthopaedics and sports medicine. To date, however, access to such datasets remains astonishingly restricted. In this CAMS-Knee project, we have directly addressed this deficit by providing accurate kinematic and kinetic data in a small population during a range of functional activities. After their public release in 2018, these unique datasets will lay the foundations for understanding the complex interactions between the hard and soft tissue structures in the human knee and can thus be used towards e.g. verifying and improving novel surgical implants, developing injury prevention or rehabilitation strategies, and analysing the biomechanics of joint degeneration, but will also importantly provide a gold standard for the validation of biomechanical computational models.

To date, the most extensive datasets that include accurate measurements of the internal tibio-femoral joint contact forces have been made available on www.orthoload.com or published as part of the grand challenge (Fregly et al., 2012; Kinney et al., 2013); but this data is limited in several ways, including number of subjects, accuracy of the kinematic measurements, extensiveness of the datasets (repetitions, number of activities, etc.), and limited range of measurement for moving activities. The data measured within the CAMS-Knee project are the first datasets to be made publicly available that include comprehensive data on multiple subjects, activities, repetitions and synchronized measurement technologies.

The CAMS-Knee data are aligned with previous reports on internal tibio-femoral loads (Fregly et al., 2012; Kinney et al., 2013; Kutzner et al., 2010), in which forces of 2.5–3 BW for normal walking were presented. The observed higher forces during stair descent and ramp descent were not unexpected due to the increase in muscle activity required to induce the movement and joint stabilisation in these more challenging exercises, and are also consistent with previous measurements in these subjects (www.orthoload.com). In terms of kinematics, it is clear that this highly constrained prosthesis limits the motion of the knee, even during walking, producing similar A-P translation and internal-external rotation to other implants (Guan et al., 2017; Mahoney et al., 2009), but slightly less rotation compared to the natural knee (Lafortune et al., 1992), where internal-external rotation of up to 10° was observed. The collected data were not all obvious however, and open a number of questions for further investigation. For example, p-d motion between the implant components was observed in the presented level walking data (Fig. 4), posing the question as to whether lift-off of the femoral component occurs. In this case, we are of the opinion that no lift-off occurs, but rather that the simultaneous posterior translation of the lateral condyle pushes the components apart due to the congruous shape of the inlay, but that contact with the inlay still remains; a hypothesis
that is supported by the smooth proximal movement of the femoral component over an extended duration of the cycle, but also that this period coincides with the phase of highest tibial internal rotation. However, it is entirely possible that the loading and unloading of the implant, which has hardly been measured throughout complete cycles in real world (non-treadmill) scenarios previously, could partially explain this anomaly. The observed peak p-d motion ranges from ca. 3-7mm in other subjects and trials, however, possibly indicates the requirement for improved 2D-3D component registration to remove outliers, or indeed possible lift-off of the femoral component. Further investigation into such aspects is clearly warranted, and will hopefully be undertaken in future collaborative projects.

A number of limitations to the quality of the data collected exist. First and foremost, only a single-plane fluoroscope was used for the analysis of internal kinematics, and 2D-3D fitting accuracy in the out-of-plane axis is known to be lower than for in-plane registration (Foresti, 2009). Consequently, the assessment of e.g. femoral component to inlay contact may be limited. In addition, the quality of the images, and therefore the accuracy of 2D-3D registration, is limited when the second leg crosses through the imaging plane. This problem was exacerbated in this study since four out of the six subjects assessed possessed bi-lateral TKAs. As a result, a small number of images were obscured by the contralateral implant and could not be reconstructed. In addition, the subjects assessed in this study averaged 68 years old and possessed at least one TKA. The interpretation and extrapolation of any musculoskeletal assessments to younger or healthy subjects may be limited, especially in light of the highly congruent INNEX knee implant that is known to considerably constrain tibio-femoral translation and rotation. Unfortunately, no detailed analysis of the pre-operative kinematics and kinetics was performed, therefore also restricting a deeper understanding of any correlation with e.g. pre-operative limb alignment. Finally, it is known that the moving fluoroscope can encumber free walking through unusual noise and visual impediments, producing a kinematic equivalent of slow walking (Hitz et al., in review). Despite these limitations, the ETH Zürich moving fluoro scope is one of the only systems available worldwide that is able to track the knee during complete cycles of activities of daily living, and thereby still offers a unique insight into joint kinematics in combination with GRF measurements, throughout loaded and unloaded phases of gait.

After a proprietary period for data analysis, the comprehensive CAMS-Knee datasets will become freely available for non-commercial usage at www.cams-knee.orthoload.com. In order to download the full datasets, recipients will be required to sign a

Fig. 4. Exemplary outcomes of some of the primary kinematic and loading datasets are shown for a single subject throughout different activities of daily living. The thick lines represent the average of all repetitions for this one subject, while the associated standard deviations are shown as a shaded area. The single maximum (*) and minimum (†) values from all subjects and all trials of normal walking are shown to provide an impression of the range and timing of the most extreme values observed across the datasets of the entire study population.
licence agreement, provide full name, position, and contact details, but also specify their intended usage of the data. With this information, we anticipate building a community of users, who will be able to interact, support each other, and even provide e.g. open source models based on the datasets. As a result, we expect the CAMS-Knee data to positively impact on current scientific and clinical approaches for the assessment and management of joint disease and injury, with tremendous potential for becoming reference datasets in medical innovation world-wide.

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Conflict of interest statement

There are no conflicts of interest.

References


