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Gait and Posture xxx xxxx

Title:

Predicted Knee Kinematics and Kinetics during Functional Activities using Optimised Motion Capture and Musculoskeletal Modelling in Healthy Older People.

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Abstract:

Knowledge of joint forces and moments is essential for comparisons between healthy people and those with pathological conditions, with observed changes at joints providing basis for a particular intervention. Currently the literature analysing both kinematics and kinetics at the knee has been limited to small samples, typically of young subjects or those who have undergone joint arthroplasty. In this study, we examined tibiofemoral joint (TFJ) kinematics and kinetics during gait, sit-stand-sit, and step-descent in 20 healthy older subjects (aged 53-79 years) using motion capture data and inverse dynamic musculoskeletal models. Mean peak distal-proximal forces in the TFJ were 3.1, 1.6, and 3.5 times body weight (N/BW) for gait, sit-stand, and step-descent respectively. There were also significant posterior-anterior forces, with sit-stand activity peaking at 1.6N/BW. Moments about the TFJ peaked at a mean of 0.07Nm/BW during the sit-stand activity. One of the most important findings of this study was variability found across the subjects, who spanned a wide age range, showing large standard deviations in all of the activities for both kinematics and kinetics. These data have provided an initial prediction for assessing kinematics and kinetics in the older population. Larger studies are needed to refine the database, in particular to reduce the variability in the results by studying sub-populations, to enable more robust comparisons between healthy and pathological TFJ kinematics and kinetics.

Keywords: Gait; Sit-Stand; Knee; Forces; Moments; Inverse Dynamics; Motion Capture

1. Introduction:

Analysis of human movement is key in order to expand the current knowledge of joint loading and mechanisms of injury and pathology. Over the years there have been many different methods to assess movement in activities of daily living (ADL), giving insight into joint kinematics and kinetics. It is known that there is variance in the way an individual will move compared to another, however the extent of this variance between age groups, genders, and ethnicities is not very well established.

External marker motion analysis remains the most widely used non-invasive method of assessing functional movements. However motion analysis has been shown to have several significant errors, with soft-tissue artefact (STA) being one of the largest [1]. Optimisation methods [2, 3] have been developed to try to reduce error associated with estimating joint kinematics from motion capture data, however to date there are no generic robust methods of accurately counteracting STA. In recent years there has been a increase in the number of commercial and freeware musculoskeletal (MS) modelling applications designed to analyse ADL from motion capture and force plate data. It has been well documented that there have been improvements made to both motion capture and MS modelling over the last 15 years [4]. The process of converting motion capture data to MS models has been shown to be very susceptible to error [5], however advances in MS modelling are aimed at making the conversion process more accurate and reliable [6, 7]. If these modelling technologies are to be used in the clinical setting further validation is needed, although the potential to predict joint loading accurately would offer valuable feedback on rehabilitative techniques and goals.

To date there are a few examples of motion capture to inverse MS modelling looking at a variety of ADL for both hip and TFJ kinematics and kinetics [8-13]. However these studies have focused on participant groups with pathology (joint arthroplasty, or neuromuscular disease) or younger individuals, and little is known about the older healthy population. Recently there has been studies based on telemetric knee prosthesis data which has given a new insight into joint reaction forces and moment [14-16], with the latest results showing previous inverse MS models may have overestimated TFJ kinetics. In the latest study using the telemetrised technology significant variance in TFJ reactions and moments during ADL were observed, peak resultant reactions deviated a whole body weight during level gait, sit-stand-sit, and stair ascent/descent between five patients [14]. However these outputs were from knee arthroplasty (KA) patients and its well known that post-operative gait can be altered from the healthy aged matched population [17]. There is a need for further investigation into TFJ kinematics and kinetics of healthy older people in order for differences between pathological findings and surgical interventions.

Gait has been the most researched activity in the current literature base of kinematics and kinetics [13, 16, 18], with a growing body of evidence analysing sit-stand and stairs [15, 19]. There is a need to investigate the range of activities in the older healthy populations, in order to gather baseline data against which to compare pathological subjects. The purposes of this study were to investigate the TFJ kinematics and kinetics in a group of older healthy volunteers during gait, sit-stand-sit, and step-descent activities, using motion capture and inverse MS modelling techniques.

2. Methods:

2.1. Participants

Twenty healthy individuals were recruited from the local community in Southampton. Institutional ethical approval and informed consent were obtained prior to data collection. Participants were excluded if they had previous lower limb pathology in the last two years, or a neuromuscular /musculoskeletal disease. Participants were aged matched to KA patients as part of a wider project (Table 1).

TABLE 1.

2.2. Instrumentation and data collection

Motion capture data during gait, sit-stand, and step-descent were taken using a 6 camera VICON 460 system (Oxford, UK), with 2 Kistler force plates (Kistler Instrument AG, Winterthur, Switzerland). Marker data were collected at 120Hz and analogue data from the force platforms at 1080Hz. Nine millimetre retroreflective markers were placed directly on the skin of each participant using double sided adhesive tape. Markers were placed in a modified Helen Hayes [20] marker set-up with anatomical landmarks established by a physiotherapist (PW). Participants were asked to walk along a 10m raised platform at a self-selected speed, and perform sit-stand-sit and step-descent activities three times.

Marker data were labelled and processed in Nexus (VICON, Oxford, UK), and exported along with the force plate data. If markers were occluded for more than 0.1 of a second, the trial was removed; this left all 20 participants with gait data but only 17 with sit-stand data (Anterior superior iliac crest occlusion during trunk flexion). Gait data were normalised to 0-100% of activity and sit-stand were normalised from full sitting to standing with knee and trunk extended. The chair used for the sit-stand activity was of a standard 45cm height and the back of the chair was removed to ensure all pelvic markers were visible to the motion capture cameras. The step-descent activity was performed from a single standardised 18cm step, beginning with the feet together at the top of the step and finishing with feet together on the floor, for which there were 18 subjects included (occlusion of heel marker). The participants selected which leg to lead with during the step-descent and performed the activity at their own self selected pace.

2.3. Modelling

Motion capture and force plate data were imported into an Inverse dynamics MS modelling software (AnyBody, Aalborg, Denmark) [21]. A baseline model of a static standing trial was taken from the VICON motion analysis system (STA is assumed to be minimal during quiet standing), and used to create the subject specific model. The 13 segment rigid body model, with 16 degrees of freedom, was orientated in the software to reflect the position of the participant being modelled.

Generically scaled models of each participant were created from an anthropometric data set from Klein-Horsmann et al. [22]. This data source was used to model mass, inertia points and muscle sites/geometry for all of the segments. The models were structured with joint centres (located according to the International Society of Biomechanics (ISB) standards [23]) and muscle attachment sites and geometries, which were scaled in accordance with a linear geometry scaling law (Equation 1).

$$s = Sp + t \qquad (1)$$

Where s is the scaled point, S is the scaling matrix, p is the original point, and t is the translation. A length, mass, fat, scaling law was used to scale soft tissues which takes into account BMI was used to scale the soft tissues in the body. When the models were scaled and positioned, the marker coordinates relative to the segments were estimated. This was accomplished by changing the location of assumed marker positions in the local coordinate frame of each of the segments.

The static models were then kept for the subsequent dynamic models. During the dynamic modelling process the kinematics were equated using a optimisation method, which defines the position of each segment in relation to the measured markers, subject to the 16 degrees of freedom in the model [2]. Rigid and optimised marker coordinates were selected depending on the known STA influence (thigh and shank markers fully optimised) [1], this method has been validated against bone pin markers for gross TFJ flexion [6] and shown to be robust under known variance in anatomical landmark definition and scaling factors [7]. Once optimised kinematics were derived, inverse dynamics was then performed with a Min/Max recruitment solver [24], with over 300 Hill Type muscles selectively recruited to solve the indeterminacy. The TFJ was modelled as a hinge joint for flexion/extension; this constraint was applied due to the known STA error [6], secondary kinematics (Anterior-Posterior and Medial-Lateral translations) were therefore not sought. The joint contact forces were taken from resultant inter-segment loading and muscle forces acting across the joint. The results presented are the joint constraint reactions, the only degree of freedom, flexion moment, is the driving moment for TFJ flexion. When this musculoskeletal modelling application was directly compared to telemetrised TFJ prosthesis data, there was a clear trend in reactions, however a over-prediction of total TFJ reactions were reported [25]. The data presented in this study should be interpreted given the known over-prediction in the modelling process.

2.4. Statistical Analysis

Resultant TFJ kinematics and kinetics from the three trials were averaged and collated for all participants. Maximal values of the constraint reactions at the TFJ from the average of the three trials are presented. Descriptive statistics are also presented as mean, range and standard deviations (SD) of each waveform of gait, sit-stand-sit, and step-descent in each figure.

3. Results:

The MS modelling of functional activities shows the variation for all the activities within the older healthy group studied.

3.1. Gait

Gait cycle parameters were output to assess the range of velocity, cadence, stride length, and double support time. The results show that there was relatively little deviation in the parameters of the individual's gait (Table 2). There were no statistical relationships between the gait cycle parameters and the magnitude of predicted forces and moments. TFJ flexion shows considerable variation (max SD = 9.77^{o}) across the gait cycle (Figure 1).

FIGURE 1.

Kinetics at the tibiofemoral joint also presented large standard deviations across the older participants. The D-P reaction forces had the largest SD values during stance phase (Figure 2), with values as high as 0.89N/BW (~565N). During gait maximal D-P reaction forces ranged from 2.72-4.35N/BW between the participants , with average stance phase taking up 63% of the gait cycle. Mean peak distal-proximal (D-P) reaction forces were 3.06N/BW (~2378N). Anterior-Posterior (A-P) reaction forces also followed the pattern previously reported, however the SDs between the participants again showed considerable variation (peak SD =0.31N/BW) through stance phase of gait (Figure 2). Varus was the largest of the internal moments across the TFJ, peaking at 0.067Nm/BW, with flexion and external rotation moment peaking at 0.041 and 0.0085Nm/BW respectively (Figure 3).

FIGURE 2&3.

3.2. Sit-Stand-Sit

TFJ flexion ranged from 101.9^{o} to 4.7^{o} during the sit-stand activity and participants all exhibited a similar TFJ flexion-extension pattern (Figure 1). Within participant right and left TFJ flexion patterns showed very similar results, with mean difference in peak extension and flexion of 1.1^{o} and 0.08^{o} respectively. TFJ flexion during the sit-stand-sit activity showed fairly high standard deviation between the participants, at the beginning and ending of the activity.

TFJ reaction force data for sit-stand activity did not reflect that of the kinematics in terms of variation, with all reaction forces and moments showing large standard deviations (Summary of the kinetic data is given in E-Appendix). D-P reaction forces at the TFJ exhibited a similar pattern for all participants, but did vary in magnitude when normalised to body weight (Max SD=0.72N/BW). PA reaction force increased sharply to 1.64N/BW during the first 18% of the activity then declines as the TFJ flexion angle decreases. Both VV and IE moment reactions showed large variance across the participants, with IE variance peaking at 18% of the cycle. Magnitudes of both VV and IE moment were higher than that of gait, with mean varus moment peaking at 0.074Nm/BW and external rotation moment peaking at 0.06Nm/BW.

3.2. Step-descent

TFJ flexion ranged from 91.1^{o} to 15^{o} during the step-descent activity, with the highest range found in the standing limb. Variance in the kinematics during step-descent was the largest of all for the activities studied, and could have been influenced by subject height, leg length and technique in descending the step. Kinetics ranged considerably (Table 2), with the highest loading in the distal-proximal direction of all the activities (mean peak = 3.46N/BW, SD = 1.42N/BW), and the highest valgus-varus moment (mean peak 0.054Nm/BW).

TABLE 2.

4. Discussion:

This study has characterised TFJ kinematics and kinetics in healthy older people during functional activities of gait, sit-stand-sit and step-descent. These data add to the current literature base for TFJ function, and have highlighted the variance found in kinematics and kinetics in the older population. When comparing the outputs of the models to the current data in the literature, they perform very favourably with the current MS modelling evidence base, however the estimated TFJ kinetics were greater than those found previously *in-vivo [16]*. This is, however, a study of older healthy individuals, and comparison with existing literature is limited due to the difference in participants and methodology from previous studies. During gait, TFJ flexion angle correlated well with previous motion capture experiments analysing a similar age group, they also exhibit similar standard deviations in the findings [26]. When comparing the D-P loading to other predictive models, such as Taylor et al [13] and Costigan et al [9], the results are very similar. Taylor et al [13] found the average peak D-P joint loading from hip arthroplasty patients to be 3.1N/BW (individual forces of 3.2, 3.2, 3.0, 2.9N/BW), and Costigan et al [9] found a mean of 3.7N/BW (SD ±1.07N/BW) respectively [9]. Higher loading results found by Costigan et al [2] could be due to the lower age of the group studied, with mean peak TFJ flexion moment of 0.54Nm/kg compared with 0.4Nm/kg in our older healthy population. The Costigan et al findings indicate a faster and potentially higher loading gait cycle in the younger subjects [2]. Total loading data from Taylor et al showed considerable variance (2.97-3.33N/BW) across the 4 tested hip arthroplasty patients assessed [13]. Taylor's data set was from a similar age group, however the history of hip arthroplasty could have an effect on joint loading. Predicted TFJ I-E moment followed a similar pattern to that found in *in-vivo* testing [16]; magnitudes of the moment complied with those seen in the telemetrised studies, with peaks of approximately 0.008Nm/BW [16].

Sit to stand data sets again compared relatively well with the limited current *in vivo* evidence base for the D-P reaction forces, with the participants' predicted D-P loading being lower than that measured *in-vivo*. Total TFJ loading during sit-stand (2.65N/BW) and stand-sit (3.32) does not, however, reflect the *in-vivo* findings of D'Lima et al (2005) after TFJ arthroplasty, who found more loading during sit-stand (2.92N/BW) than stand to sit (2.64N/BW) [27]. This difference in findings could be partly attributed to the known adaptations of sit-stand-sit activity post KA [27]. The predictive methods also showed differences in TFJ moment reaction for sit-stand. Magnitudes of I-E moment are higher in the early stages of the activity, and lower external rotation moment is seen at the end of the activity when comparing with *in-vivo* data [28]. Similar differences between predicted and *in-vivo* TFJ kinetics were found in the stand-sit data, with the P-A reaction being the much larger in the predictive results than the telemetrised data. This is mainly due to the reaction the quadriceps femoris created during the high demand activity.

The step-descent activity is difficult to compare with the literature; the closest activity to the one reported in this paper is stair descent. However it is of note that stair descent is often a reciprocal activity and the step-descent activity performed during this study was a closed chain movement. When comparing the data to other predictive models, forces appear lower than that of Costigan et al [1] (mean DP = 3.45N/BW, PA = 1.19N/BW) and Taylor et al [5] (mean DP = 5.1N/BW, PA = 1.3N/BW). However this could reflect the difference in the open chain reciprocal activity of stairs and the closed chain activity of step-descent. The results of the present study (mean D-P = 3.46N/BW, PA = 1.4N/BW), showed a marked increase in P-A reaction force when compared to the *in-vivo* shear findings of Heinlein et al (0.3N/BW) [16].

4.1 Variance and limitations of the study

The TFJ loading presented in this study varies from 2.79-4.53N/BW for the 20 subjects, which reflects the general variance found in the older healthy population, and is similar to the variance found in the latest telemetrised KA data [14]. The large standard deviations could be due to a variety of sources. It is already understood that there is considerable variation between individuals during ADL [14]. The age range of the participants (55-79 years) and sex may have added to the variability observed, however a larger study would be required to establish the affect of these variables. The variance observed in this study was not simply due to inter-individual variability, the error involved with external marker motion analysis, and the modelling assumptions must be acknowledged. One of the main modelling limitations is that the TFJ was simulated as a hinge joint (1 DOF), when it's well established that the TFJ can rotate and translate in all 6 planes. However with the known STA in thigh and shank being highest, the error would be far greater than the motion recorded for secondary kinematics [29]. It has also been well established that the process of converting *in-vivo* motion capture data to *in-vitro* MS modelling is highly sensitive to error [5]. Generic scaling laws, simplification of segments, and ignoring soft tissue structures make modelling computationally efficient, however not representative of the normal human anatomy.

The variance found in the present study should not be ignored and is needed to broaden the envelopes of data used for pre-clinical testing of devices such as a KA. This variance has been further reduced by the normalisation in this study (Body Weight). If the absolute forces were reported there would be large ranges in data. It is also of note that only three trials of each activity were recorded and additional trials could have altered the observed intra-subject variance. Rather than using one standard, for example the International Organisation for Standard (ISO) for pre-clinical testing, there needs to be approaches where known variance is applied to the TFJ representing a variety of activities. Only then will the pre-clinical testing reflect the potential loading patterns seen across the population. Even though external marker motion analysis and inverse MS modelling have their sources of error, continued development will help towards the clinical application to assess kinematics and kinetics during dynamic movement. In order for predictive modelling to be clinically relevant further validation is needed and any assumptions in the modelling process must be acknowledged.

5. Conclusions:

This data set of TFJ kinematics and kinetics in older healthy individuals highlights the magnitude and between-subject variance found during ADL. External marker motion analysis with MS modelling has been shown to be an effective method to predict TFJ kinematics and kinetics, although further validation is required for it to be used clinically. These data need to be taken into account when comparing pathological and healthy kinematics and kinetics.

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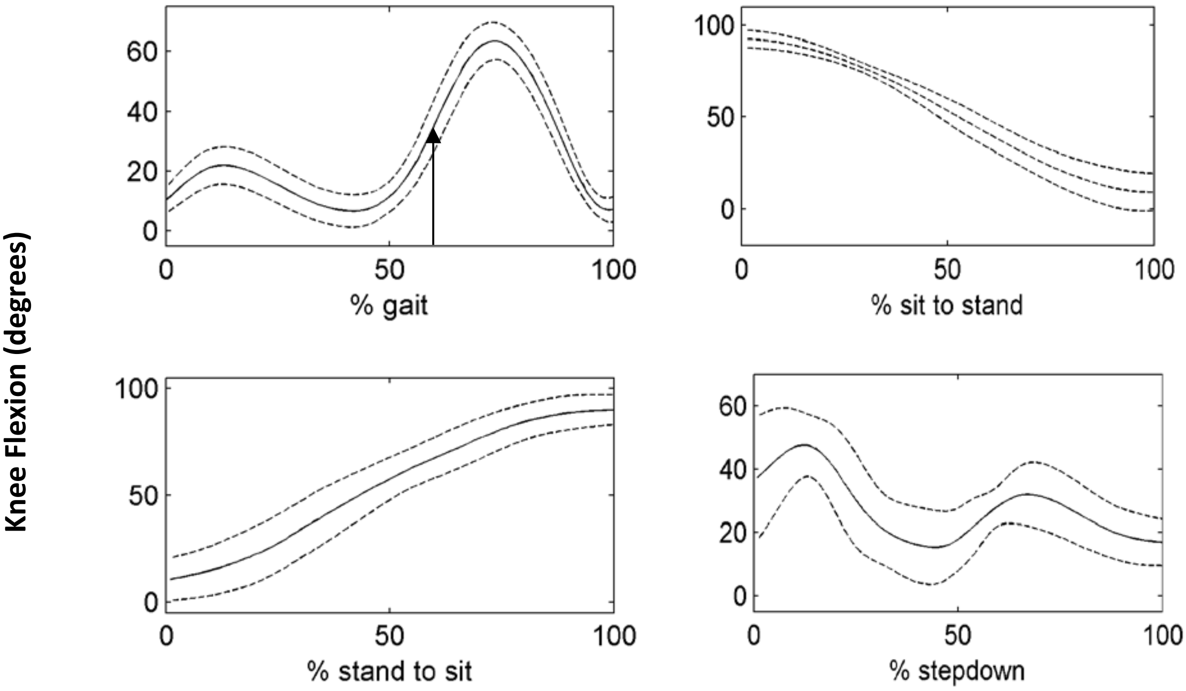
FIGURE1



Knee Flexion (degrees)
% gait
% sit to stand
% stand to sit
% stepdown

FIGURE2

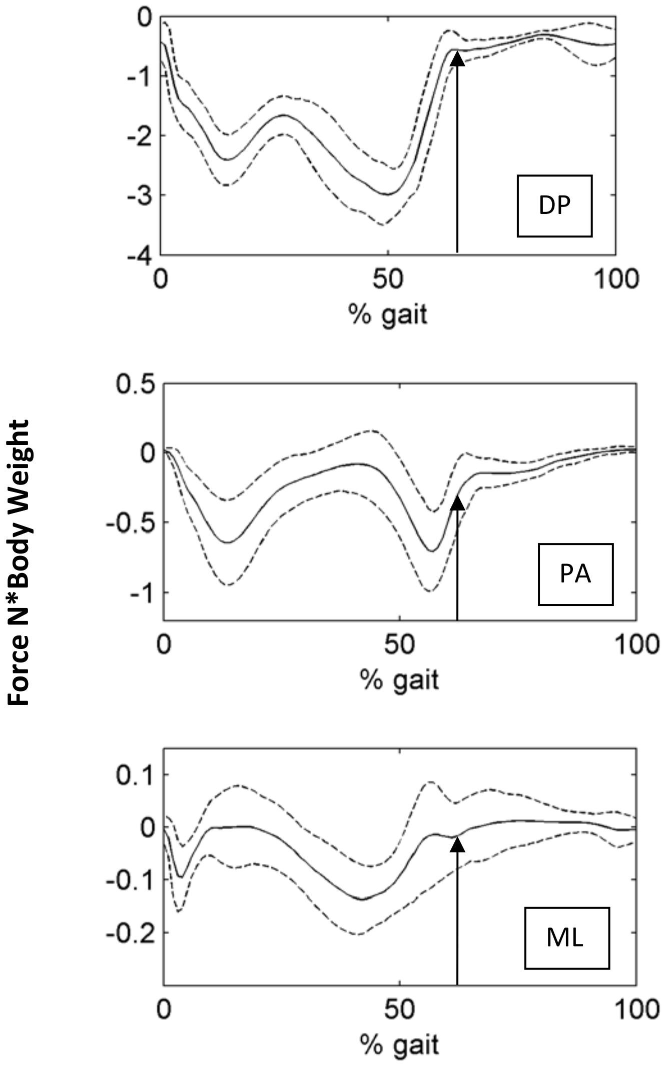
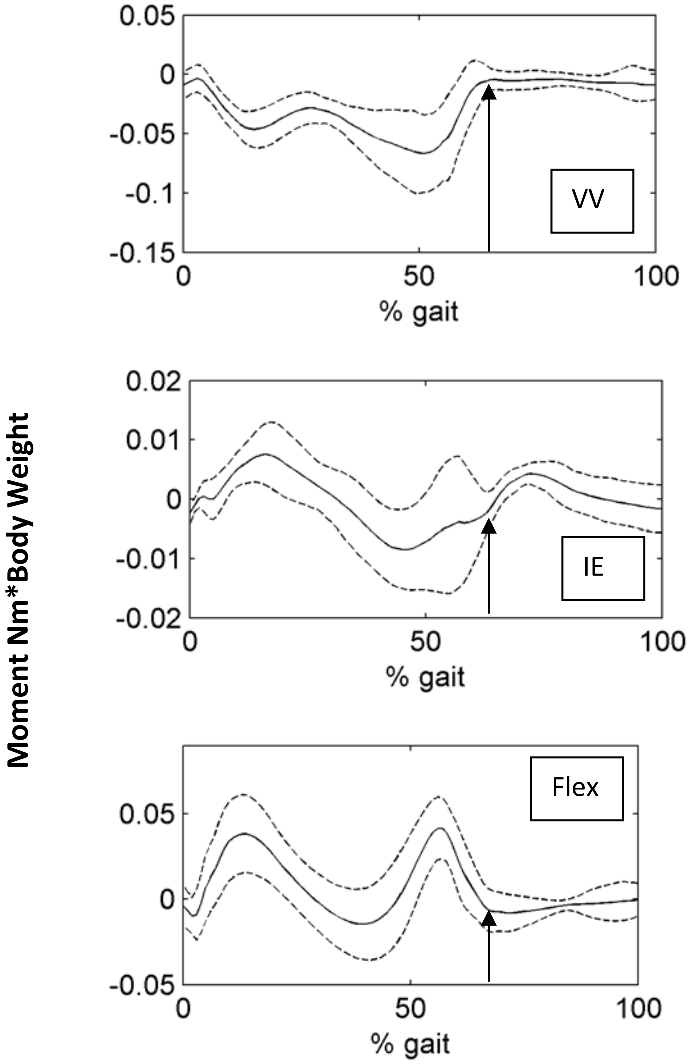


FIGURE 3.



0.05
0
-0.05
-0.1
-0.15
0
50
100
% gait
VV
0.02
0.01
0
-0.01
-0.02
0
50
100
% gait
IE
Moment Nm*Body Weight
Flex
0.05
0
-0.05
0
50
100
% gait

Table 1. Anthropometric measurements of 20 older healthy individuals.

	Age (years)	Height (cm)	Weight (kg)	BMI	Gender
Mean	62.45	1.6615	77.86	28.18	55% Female
SD	5.94	0.11	13.27	3.92	
Max	79	1.84	96	34.96	
Min	55	1.31	53	20.19	

Table 2. Mean Peak constraint reaction during three trials of level gait, sit to stand, stand to sit, and step-descent (leading leg). Distal Proximal (D-P), Anterior-Posterior (A-P), Medial Lateral (M-L), Valgus Varus moment (V-V), Internal External moment (I-E), flexion moment (Flexion). All data rounded to 2 decimal places. Gait cycles parameters are detailed, including velocity, cadence, stride length, and double support time.

KINETIC MEASURE	GAIT		SITSTAND		STANDSIT		STEP DESCENT	
	Avg.	SD	Avg.	SD	Avg.	SD	Avg.	SD
D-P (N/BW)	3.06	0.89	1.55	0.72	1.41	0.61	3.46	1.42
A-P (N/BW)	0.70	0.31	1.64	0.73	1.18	0.47	1.38	0.97
M-L (N/BW)	0.14	0.08	0.12	0.07	0.03	0.08	0.15	0.18
V-V (Nm/BW)	0.07	0.03	0.07	0.05	0.07	0.05	0.05	0.03
I-E (Nm/BW)	0.01	0.01	0.06	0.05	0.05	0.03	0.01	0.02
Flexion (Nm/BW)	0.04	0.03	0.06	0.03	0.05	0.03	0.04	0.05

Gait Cycle Parameter	Mean	SD	Range
Velocity (m/sec)	1.15	0.1	1-1.5
Cadence (steps per min)	108	7.9	95-123
Stride Length (m)	1.26	0.14	1.1-1.6
Double Support (sec)	0.24	0.03	0.16-0.28