Section: Technical Note

Article Title: Comparison of Stance Phase Knee Joint Angles and Moments Using Two Different Surface Marker Representations of the Proximal Shank in Walkers and Runners

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Abstract

Efforts to compare different surface marker configurations in 3-dimensional motion analysis are warranted as more complex and custom marker sets become more common. At the knee, different markers can been used to represent the proximal shank. Often, two anatomical markers are placed over the femoral condyles, with their midpoint defining both the distal thigh and proximal shank segment ends. However, two additional markers placed over the tibial plateaus have been used to define the proximal shank end. For this experiment, simultaneous data for both proximal shank configurations were independently collected at two separate laboratories by different investigators, with one lab capturing a walking population and the other a running population. Common discrete knee joint variables were then compared between marker sets in each population. Using the augmented marker set, peak knee flexion after weight acceptance was less (1.2-1.7°, $p<0.02$) and peak knee adduction was greater (0.7-1.4°, $p<0.001$) in both data sets. Similarly, the calculated peak knee flexion moment was less by 15-20% and internal rotation moment was greater by 11-18% ($p<0.001$). These results suggest that the calculation of knee joint mechanics are influenced by the proximal shank’s segment endpoint definition, independent of dynamic task, investigator, laboratory environment, and population in this study.

Keywords: kinematics, kinetics, gait model, marker sets

Word Count: 1649

Introduction

Numerous marker sets have been developed for three-dimensional locomotion research, varying from simpler, traditional configurations to more elaborate, custom sets.[1,2] To represent the commonly studied knee joint, often one or two markers are placed over the femoral condyles. In shared fashion, these femoral condyle markers typically help define both the distal end of the thigh segment, as well as the proximal end of the shank segment. However, shared joint marker configurations have specific model-based limitations. One consequence of a shared knee joint configuration is a less realistic representation of the proximal shank segment end, as markers over the condyles artificially lengthen the shank segment superiorly. To address this, some research groups use additional markers to independently define the distal thigh and proximal shank.[3-6] If correctly placed relative to the knee joint line, separate markers defining the proximal shank may more accurately represent the shank segment properties, as the segment would no longer be artificially lengthened. Importantly, three-dimensional knee joint angles and moments would likely be different as well. Understanding these differences is relevant to research groups participating in multicenter studies, as well as to authors looking to compare results obtained using different marker sets. Therefore, the aim of this study was to assess stance-phase changes in discrete knee joint angles and moments in walkers and runners, as well as static standing angle changes, when proximal shank markers are added.

Methods

Two data sets from different study populations (walkers and runners) were used in this experiment, and were acquired by different investigators in separate laboratories. The same methodologies were used for data collection in each lab. The two simultaneously-captured marker configurations allowed us to define the proximal shank 1) more traditionally, sharing the two markers placed over the femoral condyles, and 2) using two markers placed over the tibial plateaus below the joint line. Anatomical markers were placed over the iliac crests, greater trochanters, femoral condyles, tibial plateaus, malleoli, first and fifth metatarsal heads, and distal aspect of the shoe. Tracking markers, which remained on the subjects for the duration of testing, were placed over the anterior superior iliac spines, L5-S1 interspinous space, a cluster of three on the rearfoot, and two shell-mounted clusters of four markers over the distal posterolateral shank and thigh

Data were captured using either an eight camera Vicon (VICON, Oxford, UK) (walkers) or Motion Analysis (Motion Analysis Corporation, Santa Rosa, CA, USA) (runners) system, at a video capture rate of 120 Hz. Force data were acquired by Bertec force plates (Bertec Corporation, Columbus, OH) sampling at 1080 Hz. Pre-determined speeds were 1.5 m/s (±5%) for the walkers, and 3.7 m/s (±5%) for the runners. Five trials with complete foot strikes on the force plate surface, acceptable speed, and minimal or no marker drop-out were collected for each condition.

All trials were post-processed identically using Visual 3D (C-Motion, Germantown, MD, USA) software. The creation of the shank segment, modeled as a frustrum of a right cone, was initiated by creating the segment's frontal plane from four markers. Distally, the medial and lateral malleoli were used. Proximally, either the femoral condyle markers or the tibial plateau markers were used. These four targets were used in a least-squares fitting, such that the sum of squares distance between the four markers and the created frontal plane were minimized. The thigh was derived using a plane based on the hip joint center and the femoral condyle markers, and was identical in both processing conditions. Based on published anthropometric data, the shank was assigned 4.65% of the subject's total body mass[7]. Segment moments of inertia (IXX, IYY, IZZ) and the location of the center of segment mass were calculated per Hanavan's equations[8]. Knee joint angles (X-Y-Z Cardan sequence) were calculated using the relative orientation of the shank and thigh segments. External moments were calculated using standard inverse dynamics and expressed about the shank's proximal endpoint, the origin of that segment coordinate system. Peak knee joint angles and moments in all planes were extracted from each trial during stance for averaging using custom-written LabVIEW (National Instruments Corporation, Austin, TX) software. Comparisons between the knee joint marker configurations were conducted using paired t-tests in both task populations ($p \leq 0.05$). Static standing knee joint angles were first compared, as we expected dynamic differences may be partly attributable to changes in static pose. Dynamic joint angles were compared between marker configurations with and without normalizing the knee joint data to the standing calibration trial pose.

The first data set consisted of 15 healthy individuals (13 males, age 24.0±3.7 yrs) who performed level walking. The average height, mass, and body mass index (BMI) of the predominantly male participants were 1.74±0.08 m, 69.1±9.9 kg, and 22.7±2.2 kg/m^2 respectively. The second set (15 females, 20.7±1.2 yrs) was taken from healthy individuals while running. The average height, mass and BMI for this entirely female cohort were 1.67±0.07 m, 61.4±9.3 kg, and 22.0±1.8 kg/m^2. As such, the groups represented almost totally different gender compositions.

Results

Static differences were noted across both data sets in the sagittal and frontal planes only (Table 1). During the standing calibration trials, calculated knee angles were more extended 0.9º in the walkers and 1.6º in the runners ($p<0.001$) when the tibial plateau markers were used.

Similarly, standing adduction was greater by 1.4º in the walkers and by 0.7º in the runners ($p<0.001$).

Dynamically, peak sagittal and frontal plane angles were altered when using the tibial plateau markers. Peak knee flexion angle was lesser by 1.2º in the walkers, and by 1.6º in the runners ($p<0.05$). Peak knee adduction angle was 1.4º greater in the walkers, and 0.7º greater in the runners ($p<0.001$). In regards to peak joint moments, knee flexion moment was less by 0.07 normalized Nm in the walkers, and by 0.33 in runners ($p<0.001$). In the transverse, peak knee internal rotation moment was greater by 0.02 normalized Nm in the walkers and by 0.01 in the runners ($p<0.001$).

In a secondary follow-up analysis, when dynamic knee angles were normalized to the standing calibration, there were no differences between marker configurations.

Discussion

Previous literature has suggested that marker set variation can influence joint kinematic and kinetic data.[1,2,9-11] In our independent data sets using different locomotor tasks and populations, utilizing tibial plateau markers to define the proximal shank resulted in remarkably consistent shifts in peak knee joint angles and moments during stance. More than one mechanism is likely to contribute to these differences. One mechanism is the altered shank segment coordinate system. Due to the manner in which the shank segment is created, positionally different proximal segment markers altered the final pose of the shank's frontal plane during the static reference trial. A greater amount of knee extension and adduction during the standing trial indicates that the representation of the proximal shank segment end shifted posterior and medial relative to the shared marker configuration (Table 1). As expected, similar kinematic effects of the altered shank coordinate system are apparent in both the static and movement trials. Further, the more posterior location of the proximal shank endpoint would decrease sagittal plane knee joint moment values when the ground reaction force vector is posterior to the knee (Figures 1 & 2). The lowered sagittal joint moments occur partly because the posterior segment end shift of the shank positions the proximal shank closer to the GRF vector during locomotion, decreasing the perpendicular distance from GRF to the knee joint axis of rotation. Of lesser influence, the shank’s segment properties were also altered, as the proximal and distal endpoints were defined as the midpoints between the markers on each segment end. Therefore, although the mass assigned to the segment was constant between the two iterations of the shank, the inertial properties assigned to the segment were changed. While essentially negligible in calculating our variables of interest, we observed a 9% shorter shank segment length, a 7% inferiorly shifted segment center of mass, and slightly decreased proximal segment radius when the tibial plateau markers were used.

As expected, walking and running kinematic differences calculated using proximal tibial segment markers were minimized if they were reported relative to each participant’s standing calibration angles (commonly referred to as “normalized” joint angles). This further reaffirmed the observation that incorporation of additional markers on the tibial plateaus implements a consistent shift in peak knee angles. In principle, normalized joint angles are advantageous as they account for marker placement error and the resulting misalignment of the associated segment coordinate system. However, normalized joint angles are often not reported in favor of non-normalized kinematics, particularly when investigators who are experienced and well-trained in marker placement evaluate individuals with structural knee or hip misalignments. For example, to report normalized joint kinematics among individuals with varus gonarthrosis or excessive femoral anteversion would remove an important structural feature from knee joint frontal plane and transverse plane kinematic data. The results of this study are seemingly particularly relevant to researchers who desire to interpret gait mechanics in the context of femoral and tibial structural variability.

This study has limitations. One key limitation in this study is the use of two different subject populations for the two locomotor tasks. Specifically, the walking group was mostly comprised of males while the runners were all female. Therefore, it may be premature to generalize these findings to male runners and female walkers. Further, healthy subjects were used in this study. Therefore, caution must be exerted when interpreting these results in the context of clinical syndromes and pathologies such as patellofemoral pain and knee arthritis.

Incorporating the use of tibial plateau markers appears to consistently impact the calculation of knee joint kinematics and kinetics during walking and running. The addition of these markers appears to alter commonly extracted discrete knee joint angles and moments. We suggest that researchers who calculate knee angles and moments during walking and running be aware of these differences if considering the elimination of shared femoral and tibial segment endpoint markers.

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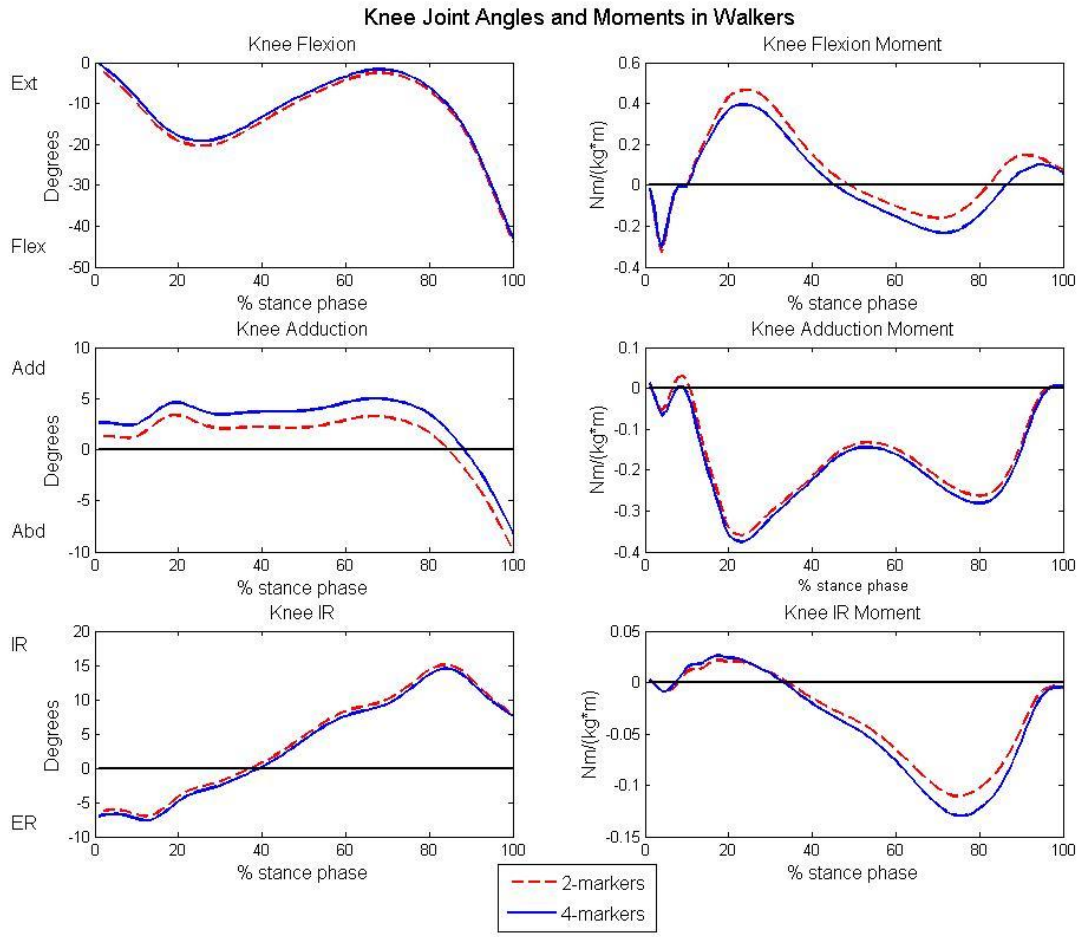

Figure 1 – Comparison of stance-phase knee angle and moment waveforms between the 2-marker and 4-marker sets in the walkers

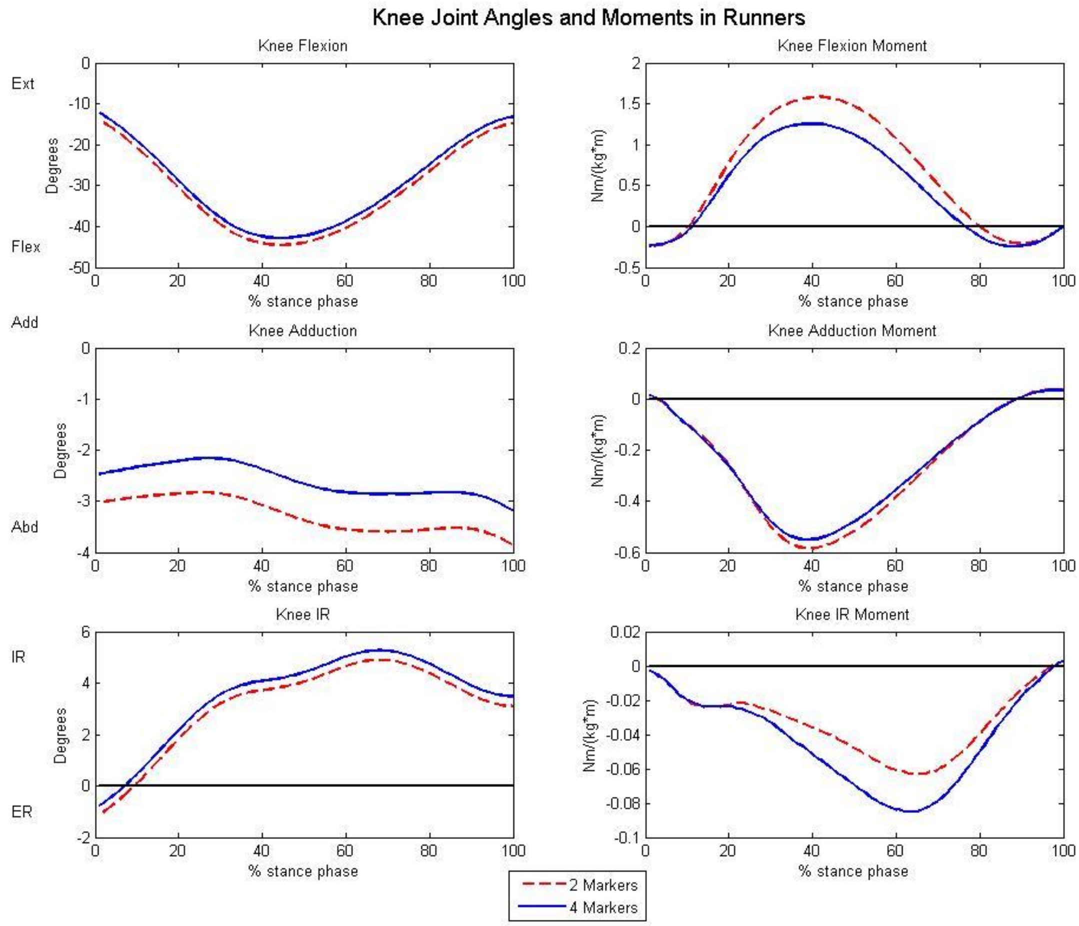

Figure 2 – Comparison of stance-phase knee angle and moment waveforms between the 2-marker and 4-marker sets in the runners

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Table 1 Means and standard deviations of standing knee angles, dynamic peak angles (normalized and non-normalized to the standing calibration angles) and moments generated using the two marker sets

WALKERS					RUNNERS				
Standing Calibration Knee Angles (°)									
	2 Markers	**4 Markers**	**Δ**	**p-value**		**2 Markers**	**4 Markers**	**Δ**	**p-value**
Extension	4.1 ± 4.5	5.0 ± 4.8	0.9	*<0.001*	**Extension**	1.5 ± 6.7	3.2 ± 6.6	1.6	*<0.001*
Adduction	4.2 ± 2.2	5.5 ± 2.2	1.3	*<0.001*	**Adduction**	-0.3 ± 2.3	0.4 ± 2.7	0.7	*<0.001*
IR	4.3 ± 3.3	3.5 ± 4.9	-0.8	0.261	**IR**	4.3 ± 7.5	4.6 ± 8.4	0.3	0.776
Peak Dynamic Knee Angles (°) (non-normalized to standing calibration)									
	2 Markers	**4 Markers**	**Δ**	**p-value**		**2 Markers**	**4 Markers**	**Δ**	**p-value**
Flexion (WA)	20.7 ± 5.9	19.5 ±5.3	-1.2	*0.022*	**Flexion (WA)**	44.7 ± 5.4	43.0 ± 5.3	-1.7	*<0.001*
Adduction (1st half)	3.8 ± 2.9	5.1 ± 3.5	1.3	*<0.001*	**Adduction (1st half)**	-1.1 ± 5.4	-0.4 ± 5.3	0.7	*<0.001*
IR (2nd half)	15.4 ± 8.4	15.1 ± 8.4	-0.3	0.819	**IR (2nd half)**	5.9 ± 7.8	6.3 ± 7.2	0.4	0.714
Peak Dynamic Knee Angles (°) (normalized to standing calibration)									
	2 Markers	**4 Markers**	**Δ**	**p-value**		**2 Markers**	**4 Markers**	**Δ**	**p-value**
Flexion (WA)	23.0 ± 8.3	23.0 ± 8.3	0	0.374	**Flexion (WA)**	46.1 ± 6.6	46.1 ± 6.6	0	0.384
Adduction (1st half)	1.1 ± 7.1	1.1 ± 7.1	0	0.187	**Adduction (1st half)**	4.5 ± 6.1	4.5 ± 6.1	0	0.924
IR (2nd half)	10.4 ± 2.6	10.4 ± 2.7	0	0.405	**IR (2nd half)**	2.3 ± 4.2	2.3 ± 4.2	0	0.543
Peak Dynamic Knee External Moments (Nm/(kg*m))									
	2 Markers	**4 Markers**	**Δ**	**p-value**		**2 Markers**	**4 Markers**	**Δ**	**p-value**
Flexion	0.48 ± 0.15	0.41 ± 0.12	-0.07	*<0.001*	**Flexion**	1.60 ± 0.26	1.27 ± 0.24	-0.33	*<0.001*
Adduction	0.38 ± 0.07	0.39 ± 0.06	0.01	0.072	**Adduction**	0.60 ± 0.27	0.56 ± 0.21	-0.04	0.167
IR	0.11 ± 0.03	0.13 ± 0.03	0.02	*<0.001*	**IR**	0.09 ± 0.06	0.10 ± 0.06	0.01	*<0.001*