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Knee moment profiles during walking: errors due to soft tissue movement of the shank and the influence of the reference coordinate system

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Abstract

The effect soft tissue movement of the shank had on knee joint moments during natural cadence walking was investigated in this study. This was examined by comparing knee moments determined from bone-anchored and surface mounted tracking targets. Six healthy adult subjects participated in this study. The largest difference (3 N m) occurred about the AP axis, with smaller differences of approximately 2 and 1 N m about the flexion/extension (F/E) and longitudinal (Long) axes, respectively. The magnitude of these differences would not likely affect the clinical interpretation of the data. The effect of reporting knee moments in two different orthogonal reference systems was also examined. The peak extension moment was significantly greater when expressed about an anatomical axis following the line of the malleoli than when the moment was reported about an axis parallel to the frontal plane of the shank. In contrast, the first peak abduction moment was significantly greater when expressed about an axis perpendicular to the frontal plane of the shank. Care should therefore be exercised whenever comparisons between studies are made in which the reference axes are not aligned. © 2002 Elsevier Science B.V. All rights reserved.

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1. Introduction

Knee joint moments during walking are common measures of instrumented gait analyses. Joint moments are determined using inverse dynamics and standard motion analysis methods. Segmental kinematics required of inverse dynamics are typically based on the motion of retro-reflective tracking targets attached to the leg. The targets are separated from the underlying bone by muscle, skin and fat, collectively referred to as soft tissue. Soft tissue of the shank moves relative to the tibia during natural cadence walking, and consequently, so do the tracking targets [1–3]. Soft tissue movement of the shank is therefore a potential source of error when estimating the net magnitude and expression of knee joint kinetics.

Only one study has reported the effect of soft tissue movement on kinetic estimates at the knee [4]. Holden and colleagues compared joint loads for three subjects during natural cadence walking based on kinematic data calculated from bone anchored and surface mounted tracking targets. Relative motion between soft tissue and bone resulted in flexion/extension (F/E) and abduction/adduction (Abd/Add) moment errors as large as 9 N m. The authors stated the magnitude of these differences would not likely affect the clinical interpretation of the data. This finding is significant since joint moments are often used in planning and evaluating patient treatment. For example, Abd/Add loading history of the knee has been used by Noyes and

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Andriacchi to help guide surgical decision making [5]. Weidenhielm et al. have shown that an excessive adduction moment in patients with medial osteoarthrosis can be reduced to normal values with high tibial osteotomy [6]. The work by Holden et al. is an important first step in documenting the effect of soft tissue movement on kinetic estimates at the knee. However, generalization of their findings must be questioned due to the limited number of subjects ($n = 3$). Moreover, the authors reported the results about an anatomically-based coordinate system in which the medio-lateral (ML) axis of the segment coordinate system (SCS) followed the line of the malleoli resulting in a SCS frontal plane that was externally rotated relative to the frontal plane of the shank.

To date, knowledge of the influence soft tissue movement has on knee kinetic estimates is limited by a small number of subjects and by results being expressed in an anatomically-based coordinate system [4]. Increased opportunities to apply this knowledge will result from expanding the data pool and evaluating the effect of expressing these influences about *traditional* functionally-based segmental axes (i.e. SCS axes parallel and perpendicular to the frontal plane of the shank). The purposes of this study therefore were to further document the effect of soft tissue movement of the shank on knee joint moments during natural cadence walking and to compare the influence of expressing these effects in functionally and anatomically-based reference systems. Furthermore, this study expands on work by Holden et al. [4] by using a surface mounted target configuration previously shown to be optimal for tracking motion of the tibia[7]. The term *optimal* is not intended to imply the tracking target configuration is the absolute best method of tracking motion of the shank, but rather it is used to describe the *best* configuration of eleven configurations tested in the aforementioned study. Errors in knee moment estimates due to soft tissue movement of the shank were expected to be smaller than values reported by Holden et al. because they did not use an optimal tracking target configuration.

2. Methods

Six adult subjects volunteered to participate in this study (one female, five male; mean age $= 25.6 \pm 1.9$ years). Five of the subjects were of average body weight and one subject was overweight based on gender, height and weight measures [8]. Average mass and height were 77.8 kg (± 14.0) and 178 cm (± 4.7), respectively. Individual subject anthropometric characteristics are reported in Table 1.

A percutaneous skeletal tracker (PST) with three spherical retro-reflective tracking targets was anchored to the medial and lateral malleoli of each subject using two modified halo pins on each side. The PST device and attached tracking targets were considered rigid with the tibia. In addition, a Velcro backed contour molded shell with four tracking targets was attached to an elasticized band wrapped around the distal lateral shank. A contour molded shell with three tracking targets was secured to the dorsum of the foot using a similar elasticized wrap. The surface mounted tracking targets (SMT) and PST targets are illustrated in Fig. 1. A detailed description of the target configurations has been described elsewhere [7].

A 6 camera, 60 Hz video based motion analysis system (VICON, Oxford Metrics, Oxford, UK) was used to collect the 3D data. Bone-anchored and surface mounted tracking targets were collected at the same time. A force platform (Bertec Corporation, Worthington, OH) was used to sample ground reaction forces at a frequency of 240 Hz. The video and force data were coincident at the video rate. The vertical force record was used to identify foot to floor contact and lift off. A 15 N threshold was used for this purpose. The interval of interest was limited to the stance phase of gait. Marker coordinates were zero phase lag filtered using a 4th order Butterworth digital filter set at a cut-off frequency of 6 Hz. Move3D (Biomechanics Laboratory, NIH, Bethesda, MD.) was used to calculate the F/E, Abd/Add and internal/external (Int/Ext) moments expressed in two differently aligned orthogonal reference systems.

Table 1
Subject characteristics

Subject	Gender	Weight (kg)	Height (cm)	Shank length (cm)	C.30%	C.50%	C.70%
1	Male	81.8	178	44.0	38.0	31.0	25.0
2	Male	79.5	180	42.0	38.0	30.0	24.0
3	Female	56.8	171	39.0	33.0	29.0	24.0
4	Male	100.0	185	42.0	40.5	32.0	25.5
5	Male	73.6	175	43.0	36.0	32.0	25.0
6	Male	75.0	178	41.0	36.0	32.0	24.0

Note: Shank length is the distance from the lateral tibial condyle (proximal end) to the lateral malleolus (distal end). C.30% = circumference of shank (cm) at 30% of shank length from proximal to distal. C.50% = circumference of shank (cm) at 50% of shank length from proximal to distal. C.70% = circumference of shank (cm) at 70% of shank length from proximal to distal.

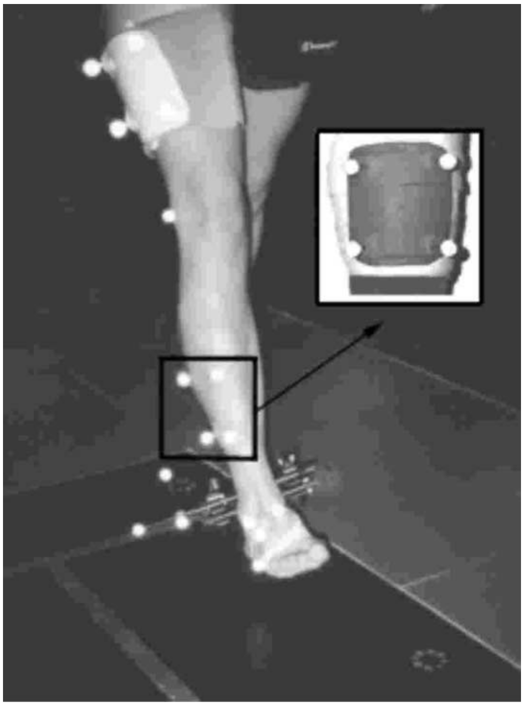

Fig. 1. Subject walking with PST device and affixed SMT tracking targets. The SMT tracking targets were positioned over the distal lateral shank, superior to the PST device. A close-up of the SMT target configuration is illustrated in the boxed region. Note, the SMT and PST tracking targets were collected at the same time. Also visible are tracking targets on the dorsum of the foot and targets attached to the thigh via a Neoprene wrap.

A segment coordinate systems for the foot and two orthogonal coordinate systems for the shank were defined from tracking targets positioned over the head of the 5th metatarsal, the ends of the fixation pins anchored to the malleoli and a target over the lateral femoral condyle. A ruler with a measurement resolution of 1 mm was used to measure the distance from the center of the targets on the end of the fixation pins to the skin over the malleoli. These distances were used by Move3D to translate the targets to the surface of the skin. The translated targets were used to define a bi-malleolar axis with the ankle joint center assumed to lie at the midpoint of the translated targets. The F/E axis of the *rotated* reference system intersected the malleoli (i.e. bi-malleolar axis). Linear offsets were applied to the translated malleoli targets, effectively rotating the distal aspect of the shank so that the F/E axis of the *neutral* reference intersected the rotated system's ankle joint center and was approximately parallel to the anatomical frontal plane of the subject. The offsets used to effect this rotation were based on subject specific amounts of distal tibial torsion as measured by an experienced physical therapist.

The longitudinal (Long) axis about which the Int/Ext moment was expressed originated at the ankle joint center and projected towards the center of the knee. The Abd/Add axis for both reference systems was defined by the cross product of the Int/Ext and F/E axes. The neutral and rotated reference systems are illustrated in Fig. 2. For the purposes of this study, knee joint moments were expressed in the SCS of the shank and calculated at the proximal end of the segment.

Joint moments were expressed in Newton-meters and time interpolated to 100 data points representing 100% of stance. Subjects walked at a self-selected pace with speed monitored by a photoelectric timing unit. Dependent t-tests were used to determine if peak extension and abduction moments using the SMT targets were significantly different when expressed in the neutral and rotated reference systems (Note: the two peaks for the abduction moment were treated separately). A Bonferroni adjustment for multiple tests was performed to reduce the probability of committing a Type I error. The adjusted alpha level for the purposes of this study was set at 0.017 (i.e. 0.05/3). Statistical comparisons for the peak Int/Ext moment were not conducted because the orientation of the Long axis was identical in both reference systems and therefore differences were not expected.

3. Results

Walking speeds for all subjects were within normal height matched walking speeds reported in the literature [9,10]. Walking speeds ranged from 1.11 to 1.36 m s^{-1}, corresponding to 1.28–1.53 s^{-1} when normalized to leg length (i.e. height of greater trochanter above floor).

A major finding of this study was that soft tissue movement of the shank during natural cadence walking had only a small effect on joint moment estimates at the knee. Differences between the SMT and PST moments reported in the neutral reference are illustrated in

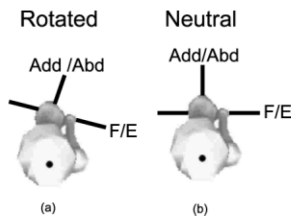

Fig. 2. Transverse view of the tibia and fibula for the rotated reference system (a) and the neutral reference system (b). The black circle in the foreground represents the knee joint center. Reference axes are displayed at the distal ends of the bones for illustration purposes. The F/E axis of the rotated reference system intersects the malleoli. The F/E axis of the neutral reference system intersects the ankle joint center of the rotated reference system and lies parallel to the anatomical frontal plane of the shank. On average, the angular difference in orientation between the rotated and neutral reference systems was approximately 15°.

Table 2
Peak knee extension moment using the SMT targets expressed in the neutral and rotated reference systems

Peak knee extension moment		
Subject	Neutral	Rotated
1	52.2	65.1
2	45.2	67.2
3	11.5	32.8
4	65.5	80.9
5	44.9	52.9
6	23.4	44.1
Mean (S.D.)	40.5 (19.7)	57.2 (17.4)
	$t=-7.306$	$P=0.001$

Moments are reported in N m. S.D. = standard deviation. Statistically significant differences in peak extension moment were noted at $P=0.017$.

Fig. 3, with similar differences noted when expressed in the rotated reference system. The curves within each plot represent differences between SMT and PST moments for each of three walking trials. Notice the scales for the F/E, Abd/Add and Int/Ext moments for individual subjects are identical, however, the range differs between subjects. The range for the ordinate was set at $\pm 2.5\%$ of the subject's mass (kg) multiplied by height (m). This scaling was selected to retain the magnitude of the difference in units of N m while normalizing the magnitude of the effect between subjects. In general, soft tissue movement of the shank had the greatest affect on the Abd/Add moment, with a peak difference of 3 N m noted for subject 1. Differences in peak moments however were not statistically significant when calculated using the SMT or PST targets. This was true for the F/E, Abd/Add and Int/Ext axes expressed in both reference systems. The magnitude of the differences ranged between 2 and 3% when reported relative to peak moments calculated using the SMT targets. These data suggest that soft tissue movement of the shank has only a small effect on knee moment estimates during natural cadence walking when an optimal SMT tracking target configuration is used.

Reporting knee moments in a neutral or rotated reference system had a significant effect on the magnitude of the F/E moment and on the magnitude and shape of the Abd/Add profile (Fig. 4). Notice how the shape of the abduction moment differs depending on the reference system in which the moments were expressed, while differences for the F/E moments were characterized by a simple DC offset. The peak extension moment reported in the rotated system was significantly greater than the peak value expressed in the neutral reference. In contrast, the first peak abduction moment was significantly greater (i.e. larger negative value) when reported in the neutral reference, with no statistically significant differences noted during the second peak. These data are reported in Tables 2–4. Joint moments based on data from the PST targets are not reported because the effect of expressing the moments in the different reference systems was similar to that for the SMT targets.

Table 3
First peak knee abduction moment using the SMT targets expressed in the neutral and rotated reference systems

First peak knee abduction moment		
Subject	Neutral	Rotated
1	−34.5	−25.5
2	−54.7	−40.7
3	−24.3	−16.3
4	−21.5	−12.6
5	−28.1	−25.4
6	−44.8	−36.1
Mean (S.D.)	−34.7 (12.9)	−26.1 (10.9)
	$t=-5.828$	$P=0.002$

The first peak occurred during the first half of stance (i.e. $<50\%$). Moments are reported in N m. S.D. = standard deviation. Statistically significant differences in peak abduction moment were noted at $P=0.017$.

4. Discussion

The purposes of this study were 3-fold. The first purpose was to expand upon the limited pool of information describing the effect soft tissue movement of the shank has on knee moment estimates during natural cadence walking. Secondly, it was a goal of this study to document the magnitude of this effect when using a SMT target configuration previously determined optimal for tracking motion of the shank. Lastly, the influence of the reference system in which the moments were expressed was examined.

Table 4
Second peak knee abduction moment using the SMT targets expressed in the neutral and rotated reference systems

Second peak knee abduction moment		
Subject	Neutral	Rotated
1	−35.0	−35.9
2	−31.9	−38.5
3	−17.4	−18.4
4	−30.7	−30.5
5	−28.8	−33.4
6	−36.5	−42.1
Mean (S.D.)	−30.1 (6.8)	−33.1 (8.3)
	$t=2.641$	$P=0.046$

The second peak occurred during the second half of stance (i.e. $>50\%$). Moments are reported in N m. S.D. = standard deviation. Differences in the peak abduction moment were not statistically significant at $P=0.017$.

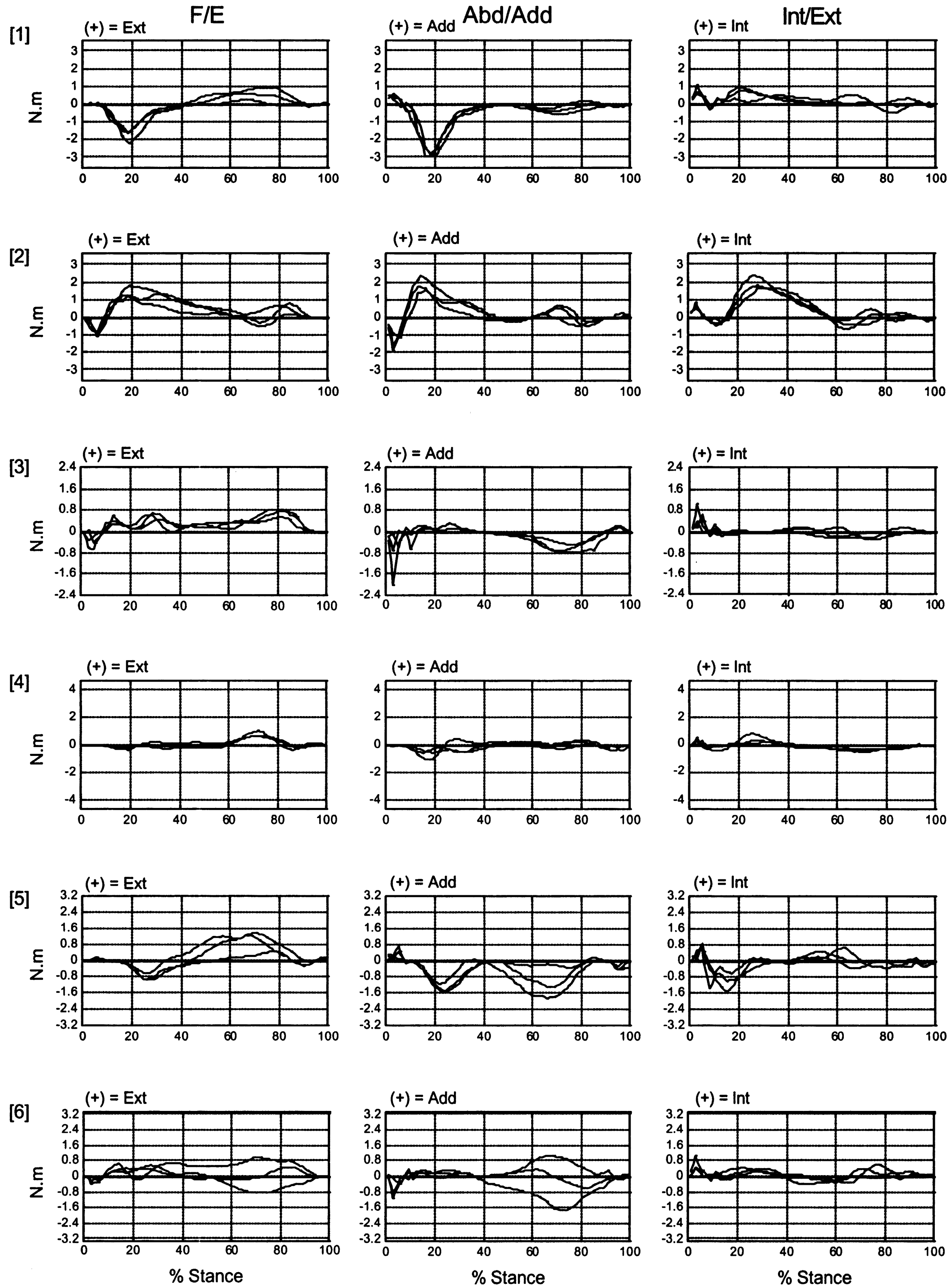

Fig. 3. Joint moment differences using the SMT and PST tracking targets (SMT–PST) expressed in the neutral reference system. The curves within each plot represent the difference between the SMT and PST moments for each of three walking trials. The number within the square brackets [ ] corresponds to the subject number reported in Table 1. Note the inter-subject variability in error profiles with the largest differences occurring about the AP axis.

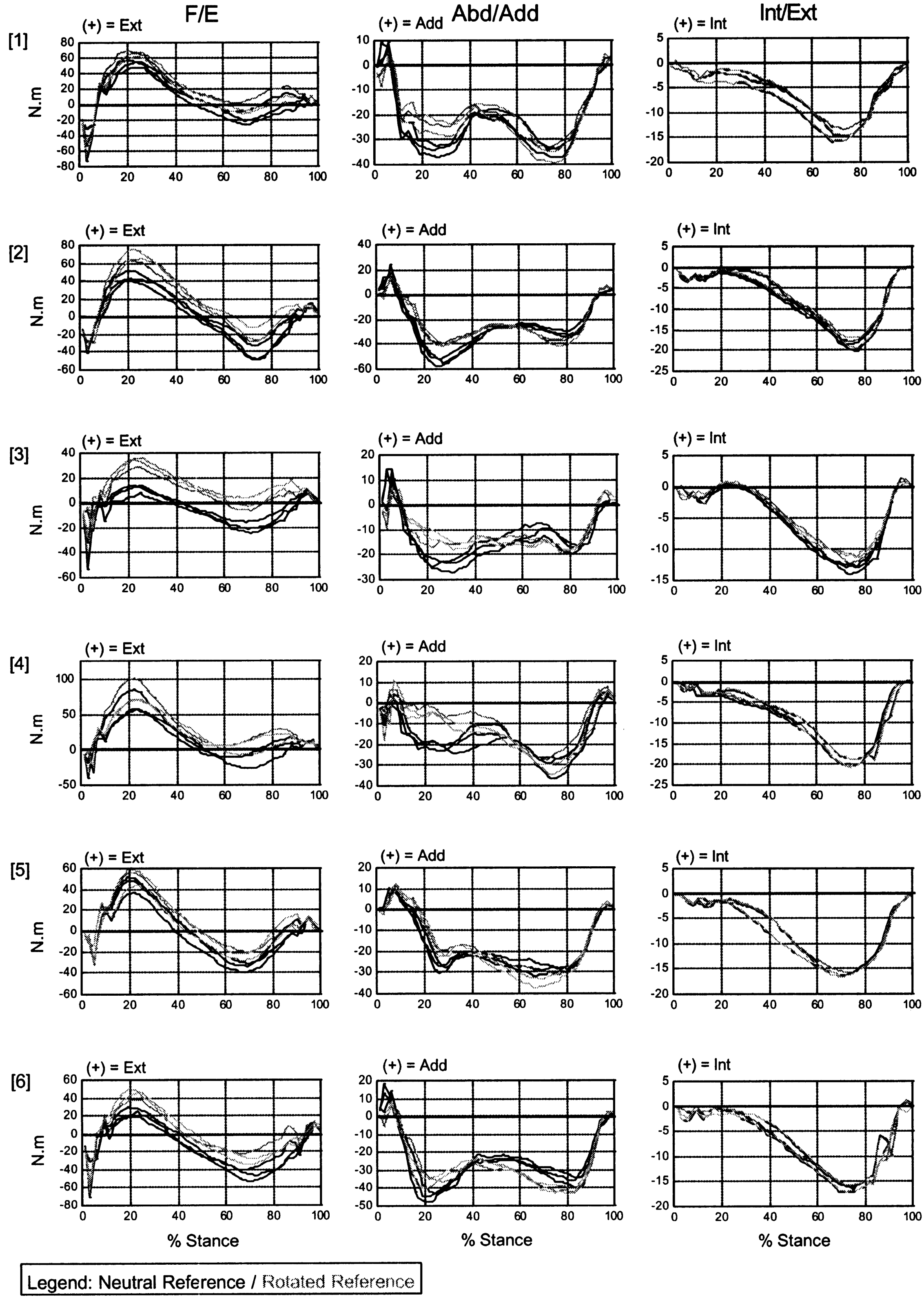

Fig. 4. Joint moments using the SMT tracking targets expressed in the neutral and rotated reference systems. The number within the square brackets [ ] corresponds to the subject number reported in Table 1. The six curves in each plot represent three trials expressed in each of the two reference systems. Note the upward shift in the extension moment when reported in the rotated reference.

With respect to the first purpose of this study, the effect of soft tissue movement of the shank on knee moment estimates has now been examined for nine subjects (Holden et al., $n=3$, Manal et al., $n=6$). Although data from both studies were not intended to characterize errors in general, both studies showed that soft tissue movement has only a small effect on moment estimates at the knee.

The errors in this study were smaller than values reported by Holden et al. [4], consistent with our expectation the differences would be smaller when using an optimal tracking target configuration. The largest difference in the present study was only 3 N m about the Abd/Add axis in contrast to 9 N m reported by Holden. Differences for the F/E (2 N m) and Int/Ext (1 N m) moments were also smaller than values previously reported (9 and 2 N m, respectively). The magnitude of these differences cannot be attributed to subjects walking at different speeds, as subjects walked at a similar pace in both studies (1.1–1.3 m s^{-1}). However, Holden et al. used a slightly different surface mounted target configuration than the optimal configuration used in this study, which may account, in part for the noted differences.

The effect of expressing the knee moments in two different orthogonal reference systems was also examined. Large differences in peak F/E and Abd/Add moments were noted for relatively small differences in the orientation of the reference axes. On average, the difference in the orientation of the reference systems was approximately 15° about the Long axis of the SCS. Peak knee extension moments were significantly greater when expressed in the rotated system ($P = 0.002$). Reporting the F/E moment in the rotated reference effectively shifted the profile upward relative to the neutral reference without affecting the shape of the curve. A similar observation is evident when comparing knee extension moments reported by Holden et al. [4] and sagittal plane moments published by Winter [11]. In contrast to knee moments reported by Winter, the profiles reported by Holden showed little or no knee flexion component. Interestingly, the orientation of the ML axis in the Holden study was similar to the orientation of the F/E axis for the rotated reference system examined here. Ostensibly, F/E moments reported by Winter were about a ML axis parallel to the frontal plane of the shank, similar to the orientation of the ML axis of the neutral reference examined in this study.

This study demonstrates that peak extension moments are significantly greater when reported about an ML axis that follows the anatomical line of the malleoli. This is a potential concern when comparing peak F/E moments between studies in which the reference axes are not aligned. Likewise, a similar concern should be raised when comparing Abd/Add moments at the knee. The magnitude of the first abduction peak (approx. 25% of stance) expressed in the neutral reference system was significantly greater (i.e. larger negative value) than when reported in the rotated reference. This finding is clinically relevant since Abd/Add joint moments during the time of the first peak are often used to guide surgical decision making and assess treatment outcomes [5,6]. In addition, the magnitude of both abduction peaks has been used to estimate medial joint forces during walking [5]. Differences in the magnitude of the second peak were not statistically different when the alpha level was adjusted for multiple comparisons. It is possible that significant differences were not detected due to a lack of statistical power related to our small sample size ($n = 6$), which is often a consequence of in-vivo studies of this nature.

The shape of the Abd/Add moment was sensitive to the reference system in which it was expressed, in contrast to differences for the F/E moments characterized a simple DC offset. That is, the magnitude of the first abduction peak was greater when reported in the neutral reference with an opposite pattern noted during the latter half of stance (Fig. 4). In addition, the timing of the second abduction peak occurred earlier when expressed in the rotated reference. Differences in the abduction moment profiles at the knee were traced back to joint reaction moments at the ankle. Defining the ML axis of the rotated reference along the line of the malleoli also rotated the proximal end of the foot and consequently influenced the expression of the ankle joint moments. This effect was carried over to the knee when the ankle moments were transformed into the SCS of the shank. Differences in the Int/Ext moment at the knee were not observed, nor were they expected because the orientation of the Long axis for both reference systems was approximately parallel at the onset.

Although the PST device was anchored to the tibia and fibula, it was assumed that relative motion between these bones would have a minimal effect on joint moment estimates at the knee. This assumption is based on work by Lundberg [12] and Svensson et al. [13] who have shown that relative motion between the tibia and fibula is approximately ± 1 mm in the lateral and posterior directions, with essentially no translation detected vertically. Translations of this magnitude were not expected to influence soft tissue movement of the shank or affect estimates of joint centers based on data from the PST targets.

The results of this study demonstrate the orientation of the reference axes should be similarly aligned when peak moment comparisons are made between studies. Another instance when the orientation of the reference system should be considered is when targets used to define the SCS are re-positioned as part of a test, re-test experimental design. It is unlikely that targets used to define the SCS can be misplaced sufficiently in a test, re-test paradigm to simulate the magnitude of the orientational differences examined in this study. The lateral and medial malleoli markers would have to be misplaced approximately 1.5 cm in the anterior and posterior directions respectively to induce the 15° difference in the orientation of the reference axes examined in this study.

In a previous paper we suggested that efforts to model relative movement between soft tissue and bone would be difficult because the method would have to account for the magnitude, the timing and the direction of the difference [7]. Likewise, modeling efforts to account for moment differences at the knee due to soft tissue movement of the shank would likely be a significant undertaking (for example, see Fig. 3, subjects 1 and 2). The effort required to model these differences however may not be warranted given the knee moment profiles using the optimal SMT targets were in close agreement with the bone based data. It is worth noting the effect of soft tissue movement reported here may represent a best case scenario, with larger differences possible had a different set of surface mounted tracking targets been used.

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