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The accuracy of estimating proximal tibial translation during natural cadence walking: bone vs. skin mounted targets

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Abstract

Objective. To assess the efficacy of estimating proximal tibial translation using video-based motion capture and an array of surface-mounted targets ideal for tracking motion of the tibia.

Design. Superficial and bone-anchored tracking targets were used to create two independent sets of data locating the proximal tibia in a global coordinate system.

Background. Knowledge of the effect that soft tissue movement has on estimates of proximal tibial translation has not been reported to date. This basic information is necessary to establish construct validity for any study proposing to document tibial translation using standard video-based motion capture methods.

Methods. A six camera motion capture system was used to collect surface-mounted and bone-anchored data for seven healthy adult subjects walking at a self-selected speed. The subjects walked along the positive Y-axis of the global reference system, with the positive Z-axis oriented vertically.

Results. Average peak differences in the location of the proximal tibia calculated from the bone and surface-mounted targets were 7.1, 3.7 and 2.1 mm along the X, Y and Z axes of the global coordinate system respectively. Individual subject peak differences were as large as 14.1, 11.8 and 8.3 mm along the X, Y and Z axes.

Conclusions. Estimates of tibial translation with a measurement resolution better than 3 mm are not likely using standard motion capture methods and tracking targets attached superficially to the lower leg.

Relevance

The results of this study clearly depict the considerable effect that soft tissue motion of the lower leg has on estimates of proximal tibial translation. Without consideration for the difficulties in measuring femoral, or patellar motion, we believe it is not feasible to routinely obtain sufficiently accurate estimates of detailed knee joint translations using superficial tracking target attachment methods.

Keywords: Soft tissue movement; Error; Knee; Motion analysis; Gait

1. Introduction

The anterior cruciate ligament (ACL) is a primary passive restraint opposing anterior translation of the tibia on the femur (Butler, 1989; Warren, 1983). A partial tear or complete rupture of the ACL can lead to joint instability often described by individuals with an ACL deficiency (ACLD) as a "giving-way" or "buckling" of the knee. A side-to-side difference in translation of greater than 2 mm is a good indicator of an ACL injury (Daniel et al., 1985). The magnitude of side-to-side differences during passive laxity tests however does not necessarily correlate well with functional stability of the ACLD knee (Snyder-Mackler et al., 1997;

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Eastlack et al., 1999). Tibial translation during dynamic tasks such as walking or jogging may prove a better predictor of functional stability since it is during dynamic activities that individuals with an ACL injury experience episodes of giving-way.

Video-based motion capture is widely used in clinical and research settings to document normal and pathological movement. Although the literature is replete with estimates of joint angular kinematics, estimates of joint translations are generally not reported. This is presumably due to an expectation that the magnitude (and possibly the direction) of the error when estimating tibial translation may be greater than the "true" translation occurring within the joint. It is important to clarify that we are not questioning the accuracy and precision of video-based motion capture systems, but rather we are concerned with the effect that soft tissue movement of the leg can have on estimates of tibial translation. Soft tissue movement is the principle source of error when estimating segmental and joint kinematics using standard video-based motion capture methods (Cappozzo et al., 1996).

Several studies have reported how tracking targets attached superficially to the shank or thigh move relative to the underlying bone during natural cadence walking (Cappozzo et al., 1996; Holden et al., 1997; Manal et al., 2000). These studies examined how *individual* targets and/or an array of targets displaced relative to the pose of a local reference calculated from tracking targets anchored to the relevant bone. Note that these data describe errors for an *individual* segment, and do not address errors in estimating translation between adjacent segments (i.e., joint translation). Reinschmidt and colleagues examined how soft tissue movement of the shank and thigh affected estimates of knee joint angular position, but they did not report errors for translation (Reinschmidt et al., 1997). Relative motion between surface-mounted and bone-anchored targets was also examined by Fuller et al. (1997). However, they did not report errors in estimating knee joint translation.

In a recent study, Ramsey and colleagues reported anterior translation of the tibia for ACLD subjects asked to perform a single leg hop test for distance (Ramsey et al., 2001). Their data were free of errors caused by soft tissue movement because the tracking targets were affixed to Steinman pins anchored to the tibia and femur. Anterior translation of the tibia on the femur ranged between 2 and 9 mm. Although it is difficult to extrapolate the results of their study to walking, their findings demonstrate that estimates of knee translation for individuals with an ACL deficiency may require a measurement resolution of only a couple of millimeters.

The magnitude of the effect that soft tissue movement has on estimates of tibial translation during natural cadence walking has not yet been resolved. The purpose of this study was to investigate the accuracy of locating the proximal tibia using tracking targets attached superficially to the shank. Note, the effect of soft tissue movement of the thigh is not accounted for in this study. Consequently, the data reported herein represent a lower bound, or a conservative estimate of the error that might be expected when estimating knee joint translation. This basic information is necessary to establish construct validity for any study proposing to quantify tibial translation using standard motion capture methods. Experimental hypotheses were not formulated per se since the purpose of this study was descriptive in nature. However, since errors in rotation and translation are not independent, and because errors in rotational estimates are smallest during mid-stance (Manal et al., 2000), it was expected that errors in locating the proximal tibia would also be smaller during mid-stance than at other times during stance.

2. Methods

Seven healthy adult subjects participated in this study. All subjects signed an informed consent approved by the Institutional Review Board at the University of Delaware. Average subject mass and height were 77.8 kg (SD 14.0) and 178 cm (SD 4.7) respectively. Individual subject anthropometric characteristics are reported in Table 1.

Two independent tracking target configurations were used to track motion of the proximal tibia in a global coordinate system (GCS). One set of tracking targets was anchored to the distal tibia using a percutaneous skeletal tracker (PST). The PST was anchored to the medial and lateral malleoli of each subject using two modified halo pins on each side. Three spherical retroreflective tracking targets were then secured to the PST device. The PST device and attached tracking targets were assumed rigid with the tibia. A complete description of the PST device and attachment method can be found elsewhere (Holden et al., 1997). A second array of four targets mounted to a Velcro backed contour molded shell was attached superficially to the distal lateral shank using an elasticized band (i.e., surface-mounted).

Table 1
Subject characteristics

Subject	Gender	Mass (kg)	Height (cm)
1	Female	61.4	169.0
2	Male	81.8	178.0
3	Male	79.5	180.0
4	Female	56.8	171.0
5	Male	100.0	185.0
6	Male	73.6	175.0
7	Male	75.0	178.0

The distal lateral location and attachment method for the surface-mounted target array has been shown to be ideal for tracking motion of the tibia during walking (Manal et al., 2000). Data for the surface-mounted and the bone-anchored PST targets were collected at the same time. Differences in the location of the proximal tibia were attributed to soft tissue movement of the shank (i.e., error). The instrumented leg of a test subject is illustrated in Fig. 1.

A six camera Vicon 370 motion capture system (Oxford Metrics Ltd., Oxford, UK) and a Bertec force platform (Bertec Corp., Columbus, OH, USA) were used to collect target trajectories and force records for each subject walking at a self-selected speed. The video and force plate data were synchronized at 60 Hz. A 15 N threshold in the vertical force record was used to identify foot-to-floor contact and lift-off. The data were time interpolated to 100 data points corresponding to 100% of stance. Move3D (Biomechanics Laboratory, NIH, Bethesda, MD, USA) was used to calculate segmental and joint kinematics from the filtered target trajectories. The X, Y and Z coordinates for each trajectory were filtered using a 4th order, zero-lag Butterworth digital filter set at a cut-off frequency of 6 Hz.

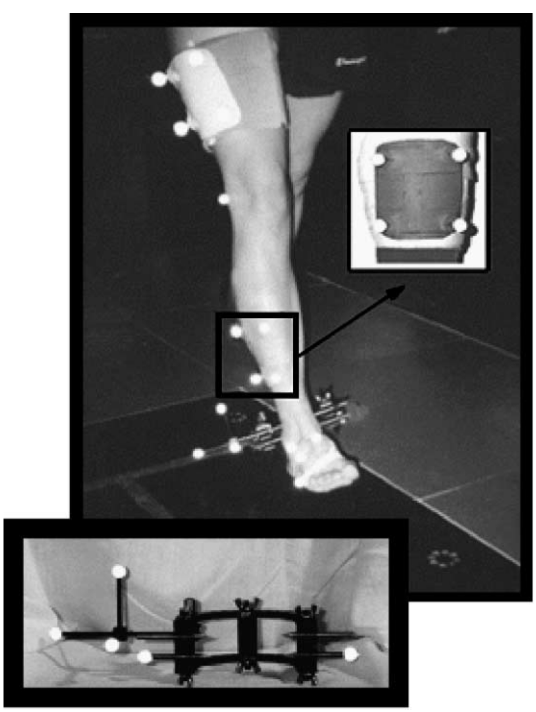

Fig. 1. Subject walking with the PST device and tracking targets mounted superficially to the leg. The surface-mounted tracking targets were positioned over the distal lateral shank, superior to the PST device. A close-up of the surface-mounted target configuration is illustrated in the inset-boxed region. A close-up of the PST device with attached tracking targets is presented in foreground-boxed region. Note the two tracking targets attached to the ends of the medial and lateral malleoli fixation pins.

The criterion measure in this study was the difference in the position of the proximal tibia estimated using the two different tracking target configurations (i.e., surface-mounted and bone-anchored). The bone-anchored location was subtracted from the surface-mounted estimate. The first step in defining the proximal tibia was to construct an anatomical coordinate system (ACS) for the shank. The ACS was defined using retro-reflective targets positioned over the ends of the fixation pins (one medial pin and one lateral pin) and a target over the lateral femoral condyle. Three-dimensional coordinates for these anatomical targets were acquired during a standing reference trial. 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 attached to the ends of the fixation pins to the surface of the skin over each malleoli. The ankle joint center was defined to lie at the mid-point between the translated "virtual" targets. Placement of the tracking targets and measurement of all distances was conducted by the same experimenter.

The frontal plane for the shank was defined using the targets over the lateral femoral condyle and the "virtual" targets over the malleoli. The positive Z-axis for the ACS of the shank was directed from the center of the ankle joint towards the knee joint center. The knee joint center was set at one half the knee diameter, medial to the target over the lateral femoral condyle and constrained to lie in the frontal plane. Calipers with a measurement resolution of 1 mm were used to measure the diameter of the knee at the widest point of the femoral condyles. The radius of the target over the lateral condyle and the length of the stem and base to which the target was attached was taken into account when defining the center of the knee. The origin of the ACS was set along the Z-axis at the estimated center of mass of the geometrical model (Winter, 1990). The shank was modeled as a frustum of a cone. The X-axis was in the frontal plane and projected laterally from the origin (i.e., to the right for a right leg). The Y-axis was determined by the right hand rule. The proximal end of the tibia and the previously defined knee joint center were defined to be coincident for the purposes of this study. The orientation of the ACS for the shank and the GCS were approximately aligned during the standing reference trial. The subjects walked along the positive Y axis of the GCS with the Z axis oriented vertically. Thus, differences in the location of the proximal tibia along the Y-axis of the GCS most closely approximate the shank's contribution to errors in estimating anterior translation of the tibia with respect to the femur. Three natural cadence walking trials were processed for each subject. Data were collected for the right leg only.

Bone-anchored data for the thigh were not collected as part of this study. Errors in estimating tibial translation caused by soft tissue movement of the shank were assessed indirectly by examining the magnitude of the errors in locating the proximal tibia using the tracking targets attached to the lower leg. It was assumed that accurate estimates of tibial translation were unlikely when errors due to soft tissue movement of the shank were large. For the purposes of this study, a large error was operationally defined to be 3 mm or greater. Three millimeters is commonly used in a clinical setting when assessing side-to-side differences in joint laxity and was approximately the average of the anterior translations reported by Ramsey et al. (2001).

3. Results

Differences in the location of the proximal tibia estimated using the surface-mounted and bone-anchored targets are presented in Fig. 2. A value of 0 indicates perfect agreement between the surface-mounted and the bone-anchored estimates (i.e., no error). Differences along the Y axis of the GCS approximate the shank's contribution to errors in estimating anterior translation of the tibia with respect to the femur, especially during mid-stance when the lower leg was approximately parallel with the Z axis of the GCS. Also included in Fig. 2 are ± 1.5 and ± 3.0 mm error bands depicted by the white and gray regions centered about the 0 mm error reference line. The white band spans the 3 mm error considered acceptable for the purposes of this study. The larger gray region (i.e., ± 3 mm) defines a boundary outside of which the magnitude of the error caused by soft tissue movement of the shank may mask what is occuring within the joint. Errors attributed to soft tissue movement of the shank were larger during the first and last thirds of stance than errors during mid-stance (Fig. 2). Absolute peak differences along the X, Y and Z axes of the GCS for the group averaged data were 7.4, 3.7 and 2.1 mm respectively. Note that the ensemble-averaged curves tended to mask the magnitude of individual subject errors. This occurred because the errors were positive in direction for some subjects while negative in direction for others; so the errors tended to cancel when averaged. For example, the ensemble average error along the Y axis during mid-stance (i.e., the thick line) was within the ± 1.5 mm error band, however, peak errors for four of the seven subjects fell outside this range, with errors for two subjects as large as 5 mm (Fig. 2). Absolute peak differences for individual subjects were 14.1, 11.8 and 8.3 mm along the X, Y and Z axes respectively.

Fig. 3 provides a functional reference relating the timing of peak errors in estimating the location of the proximal tibia as a function of knee joint angle. Smaller errors during mid-stance were associated with the knee in an extended position (see Figs. 2 and 3).

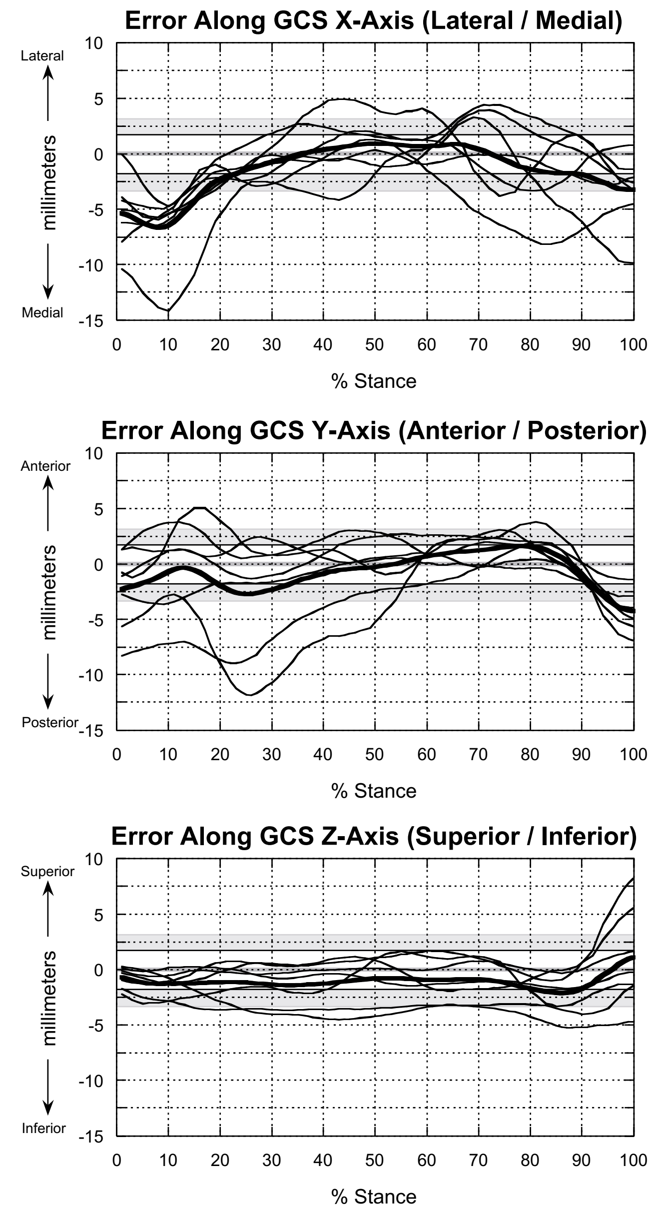

Fig. 2. Average differences in the GCS location of the proximal tibia estimated using the surface-mounted and bone-anchored targets. Three trials were averaged per subject. The bone-anchored locations were subtracted from the surface-mounted target estimates. The thin curves within each plot represent the average difference for each subject, while the thick line is an ensemble average for all seven subjects. The white and gray regions centered about the 0 mm error reference line represent ± 1.5 and ± 3.0 mm error bands. Note that the labels used to describe the direction of the error are not meant to imply anatomical translations, but rather the labels define the direction of the error for a right leg expressed in the GCS.

4. Discussion

The purpose of this study was to assess the efficacy of estimating tibial translation during natural cadence walking using video-based motion capture methods and tracking targets attached superficially to the lower leg.

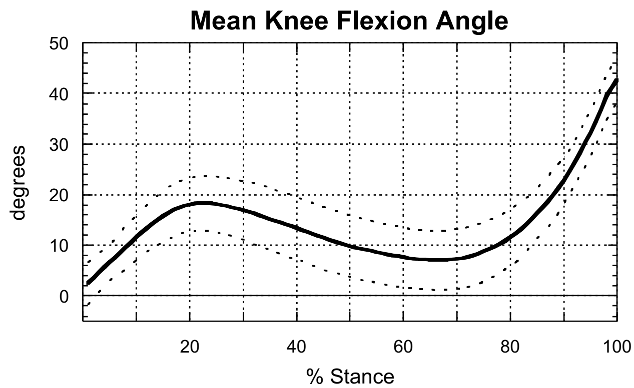

Fig. 3. Average knee flexion angle (thick line) and ±1 standard deviation (dashed lines) for all seven subjects. The data are reported as a percentage of stance phase. The surface-mounted tracking targets attached to the shank and thigh were used to calculate the knee flexion angle.

In a previous study we reported that soft tissue movement of the shank had a greater effect on angular estimates during the first and last thirds of stance than during mid-stance (Manal et al., 2000). Given that errors in translation are not independent from errors in resolving orientation, our a priori expectation was that differences in the position of the proximal tibia would be smaller during mid-stance. In general, this was noted as seen in Fig. 2. The magnitude of individual subject errors followed the knee flexion profile, with greater errors noted during periods of increasing flexion corresponding to weight acceptance during the first third and preparation for push-off in the last third of stance.

In a classic study, Lafortune and colleagues examined in vivo kinematics of the knee during natural cadence walking (Lafortune et al., 1992). The average translation of the tibia on the femur during stance was approximately 6 mm. Bone-anchored tracking argets were used in their study, and therefore the kinematic measures were free of errors caused by soft tissue movement. Smaller excursions (3 mm) were reported by Ramsey and colleagues for ACLD subjects performing a single leg hop test for distance (Ramsey et al., 2001). It is interesting to note that passive laxity measured using joint arthrometry for the ACLD subjects in the Ramsey study was greater than translations recorded during the dynamic jumping task. This finding suggests that the subjects "constrained" or "controlled" the amount of available translation. Several studies have shown that individuals with an ACL injury activate their hamstrings differently than age-matched controls (Rudolph et al., 2001; Ciccotti et al., 1994; Kalund et al., 1990). For this reason co-contraction of the hamstrings has been cited as a potential stabilizing strategy; limiting translation of the tibia on the femur. The data reported by Ramsey are important in the context of this study because they demonstrate the exacting measurement resolution that may be required to document tibial translation for individuals with an ACL injury. Errors along the X and Y axes in the present study often exceeded the translations reported by Ramsey and colleagues and occasionally the errors were larger than bone-on-bone excursions reported by Lafortune and colleagues.

In the Lafortune study, anterior/posterior translation of the tibia relative to the femur closely resembled the pattern of knee flexion/extension. Peak anterior translation occurred at approximately the same time as maximum knee extension. If bone-on-bone translations for the subjects in the present study followed a similar pattern reported by Lafortune and colleagues, peak anterior translation would have occurred at approximately 60–65% of stance when the knee was in its most extended position (Fig. 3). During this time (i.e., mid-stance) errors due to soft tissue movement of the shank were smaller than errors at other times during stance. This error however did not include contributions from the thigh, which has been shown to have a greater effect on rotational estimates than soft tissue movement of the shank (Reinschmidt et al., 1997). Thus, even during mid-stance, when errors were the smallest, soft tissue movement of the leg may "mask" what is occuring within the joint (i.e., actual bone-on-bone translation).

Soft tissue movement of the lower leg can affect estimates of proximal tibial translation in one of two different ways. First, soft tissue movement is a source of error if it causes a deformation in the target configuration (i.e., the inter-target distances change). This was not believed to have affected the results of the present study because the tracking targets attached superficially to the leg were mounted to a contour molded shell and therefore relative motion between the targets was constrained. A second source of error occurs when the tracking targets and the soft tissue to which they are attached moves relative to the underlying bone. Relative motion of this sort was believed to have been the primary source of error in the present study. Numerical algorithms based on deformation models (Cheze et al., 1995; Alexander and Andriacchi, 2001) cannot account for this type of error, and the use of an over-determined target configuration (Spoor and Veldpaus, 1980; Soderkvist and Wedin, 1993) will do little more than reduce "system noise" when the targets are mounted to a rigid shell. An appealing solution would be to filter out the contaminating soft tissue movement. Unfortunately, this is not possible because the frequency content of the soft tissue moving relative to the bones lies within the same spectra as the motion of the bones (Fuller et al., 1997; Kaufman et al., 1991). Our finding that the tibial translation results revealed no regular pattern of soft tissue error between subjects indicates the unlikely success of numerical methods for modeling and removing the soft tissue motion artifact. Accurate measures of tibial translation during dynamic activities such as walking can be made using invasive methods (e.g., PST or Steinman pins), however, novel imaging techniques would be needed to do so non-invasively.

The results of this study clearly depict the considerable effect of soft tissue motion on estimates of proximal tibial translation. It is important to note that these estimates were obtained under ideal measurement conditions and therefore could be considered minimal "best case" examples. In this study, subjects were all healthy, college-age individuals, who upon visual inspection, had minimal sub-cutaneous adipose tissue in the surface target attachment region. An ideal surface mounted tracking target method that has been previously shown to minimize soft tissue movement errors was used to track the shank segment. The motion capture camera configuration was customized to ensure the utilization of six cameras in the reconstruction of each target trajectory. Also, the spatial resolution of the motion capture system was optimized by substantially reducing the imaged volume size from that required for typical clinical gait studies.

This is the first study to report how soft tissue movement of the lower leg can affect estimates of proximal tibial translation. It is hoped that these data will be used to establish a range or a lower boundary on the error, inside of which one can not resolve if measured translations represent true bone-on-bone motion or were the effect of measurement inaccuracies caused by soft tissue movement. Without consideration for the difficulties in measuring femoral, or patellar motion, we believe it is not feasible to routinely obtain sufficiently accurate estimates of detailed knee joint translations using superficial tracking target attachment methods.

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