Computers in Biology and Medicine 92 (2018) 118–127

Contents lists available at ScienceDirect

Computers in Biology and Medicine

journal homepage: www.elsevier.com/locate/compbiomed

Model-based tracking of the bones of the foot: A biplane fluoroscopy validation study

Joseph M. Iaquinto [a,b], Matthew W. Kindig [a], David R. Haynor [c], QuocBao Vu [b], Nathan Pepin [b], Richard Tsai [a], Bruce J. Sangeorzan [a,d], William R. Ledoux [a,b,d,*]

[a] RR&D Center for Limb Loss and MoBility, Department of Veterans Affairs, Seattle, WA 98108, USA
[b] Department of Mechanical Engineering, University of Washington, Seattle, WA 98195, USA
[c] Department of Radiology, University of Washington, Seattle, WA 98195, USA
[d] Department of Orthopaedics & Sports Medicine, University of Washington, Seattle, WA 98195, USA

ARTICLE INFO

Keywords:
Biplane fluoroscopy
Model-based
Bone-based
Validation
Foot and ankle

ABSTRACT

Measuring foot kinematics using optical motion capture is technically challenging due to the depth of the talus, small bone size, and soft tissue artifact. We present a validation of our biplane X-ray system, demonstrating its accuracy in tracking the foot bones directly.

Using an experimental linear/rotary stage we imaged pairs of tali, calcanei, and first metatarsals, with embedded beads, through 30 poses. Model- and bead-based algorithms were employed for semi-automatic tracking. Translational and rotational poses were compared to the experimental stage (a reference standard) to determine registration performance.

For each bone, 10 frames per pose were analyzed. Model-based: The resulting overall translational bias of the six bones was 0.058 mm with a precision of ± 0.049 mm. The overall rotational bias of the six bones was 0.42° with a precision of ± 0.41°. Bead-based: the overall translational bias was 0.037 mm with a precision of ± 0.032 mm and for rotation was 0.29° with a precision of ± 0.26°.

We validated the accuracy of our system to determine the spatial position and orientation of isolated foot bones, including the talus, calcaneus, and first metatarsal over a range of quasi-static poses. Although the accuracy of dynamic motion was not assessed, use of an experimental stage establishes a reference standard.

1. Introduction

The lower extremity joints (hip, knee, and foot/ankle) play a primary role in locomotion and mobility but studying their function can be challenging. Commonly, optical motion capture systems use reflective surface markers and infrared light to track body segments. Optical motion capture has been used to study the lower extremity with numerous marker models [1–5]. These have been used to quantify gait kinematics in normal subjects [6,7], in patients with ankle arthrosis [8], and in patients with adult-acquired flatfoot deformity [9].

An overall limitation of optical motion capture is soft tissue artifact – the error associated with the non-rigid motion between the skin-mounted location of an optical marker and the underlying bony landmark it is nominally tracking [10–16]. While soft tissue artifact can be minimized in cadaveric studies, where markers can be directly attached to bones of interest, this method is generally unavailable during *in vivo* observation. Tranberg and Karlsson measured soft tissue artifact using metal markers and a fluoroscope. They found that marker movement was dependent on marker location – distal forefoot markers demonstrated less motion (a maximum of 1.8 mm) than proximal hind- and midfoot markers (a maximum of 4.3 mm) [11]. Another study compared bone pins to two skin- and plate-mounted markers; contrasting any two of the three protocols during stance showed an average maximum difference in error that was >3° in 100% of the data, >5° in 73% of the data and >8° in 23% of the data [15].

Two additional limitations arise when considering the foot. First, the talus possesses no near-surface landmarks due to its depth; this renders it unsuitable for optical motion capture and thus prevents the separation of ankle and subtalar joint motions. Second, many of the bones are very small, and unable to receive an adequate number of markers without experiencing significant marker visual overlap. This requires grouping into multi-bone segments, necessitating a simplification of the kinematics

* Corresponding author. VA Puget Sound, ms 151, 1660 S. Columbian Way, Seattle, WA 98108, USA.
E-mail address: wrledoux@uw.edu (W.R. Ledoux).

https://doi.org/10.1016/j.compbiomed.2017.11.006
Received 15 April 2017; Received in revised form 6 November 2017; Accepted 7 November 2017

0010-4825/Published by Elsevier Ltd.

from bones to regions.

Biplane fluoroscopy (dynamic stereo X-ray [16] or dual-orthogonal fluoroscopy [17]) is a technique which can track the motion of bones directly, thus overcoming the deficiencies of optical motion capture, including soft tissue artifact, as well as deep and/or small bones. The major drawback that is added is radiation exposure.

Biplane fluoroscopy has been used to study the hip [18–20], knee [17,21–24] and ankle [25–29]. This technology is relatively novel, has no commercially available software, and the performance of these systems may be influenced by the anatomy (large vs. small bones). Due to these considerations, there is a clear need to validate biplane fluoroscopy techniques on the anatomy of interest. Numerous laboratories have described validation methods [19,22–24,27,29–35]; most studies use bead-based (i.e., marker-based) tracking as a reference for their bone-based (i.e., model-based, the term adopted in this paper) results, where the pose of the beads is the "gold standard". At the hip, Lin et al. reported a bias ± precision of 0.60 ± 0.75 mm and 0.69 ± 0.85° in static poses [19]. Anderst et al. tracked the femur and tibia with static bias ranging from −0.37 mm to 0.14 mm and precision ranging from 0.03 mm to 0.08 mm [22]. Kapron et al. reported marker based tracking of the hip with an average bias of 0.32 ± 0.08 mm for the pelvis and 0.30 ± 0.06 mm for the femur; and with an average precision for the pelvis and femur of 0.13 ± 0.03 mm and 0.14 ± 0.04 [20]. At the knee, Bey et al. reported static values for the patella and femur of a bias from −0.174 mm to 0.248 mm, and a precision from 0.023 mm to 0.062 mm [30]. Stentz-Olesen et al. reported maximum mean errors of 0.62 mm for translation and 0.96° for rotations in the tibia and femur [23]. At the ankle, Caputo et al. determined an average error in displacement of 0.04 ± 0.11 mm and in rotation of 0.2 ± 0.1° [26]. Wang et al. reported a mean translational bias of 0.03 mm ± 0.35 mm and a mean rotational bias of 0.25 ± 0.81° across all trials and for all bones (tibia, talus and calcaneus) [27]. Cross et al. reported overall RMS error for their model-based tracking method which averaged 0.43 ± 0.22 mm and 0.66 ± 0.43° for static and 0.59 ± 0.10 mm and 0.71 ± 0.12° for dynamic trials [28].

Our laboratory has developed a biplane fluoroscopy system (referred to as a biplane system) to study foot and ankle kinematics. We have previously reported the results of our hardware optimized for marker-based tracking using a high-accuracy experimental translation/rotation stage [36].

There are biplane fluoroscopy approaches that use manual alignment of bones by human operators; this commonly involves a user matching edges in the fluoroscopic image to contours from a computed tomography (CT) model of the bone [19]. Our method uses a computational bone registration method to determine the optimal bone pose. This is similar in form to the methods used by other groups, and can take advantage of both edge and content based information from the associated CT scan [31].

Our objective here was to develop and validate a model-based tracking technique applied to the bones of the foot. Due to the complex anatomy, isolated bones (the talus, calcaneus, and first metatarsal) were selected. These frequently studied bones represent diverse shapes. A secondary objective was to evaluate marker-based tracking performance. Both objectives utilize an experimental stage as a "gold standard". With the use of a numerical optimization algorithm, we hypothesized that our system could track the position and rotation of bones of the foot with sub-millimeter and sub-degree bias and precision, during imaging performed with the same framerate and exposure time used during live subject gait trials.

2. Methods

2.1. System overview

Our biplane system works by combining a) a pair of 2-dimensional (2D) X-ray images of a subject's bones during a functional task, b) 3-dimensional (3D) models of the bones, extracted from CT and c) a virtual representation of the X-ray system geometry. The 3D models of the bones are used to mathematically generate X-ray images in multiple poses until they "match" the 2D images taken during subject trials. This methodology yields a 3D bone pose for each frame, which can then be used to calculate joint kinematics during a dynamic task. More detail is included in the bone model pose optimization section below.

Our biplane system hardware consists of two modified Philips BV Pulsera C-arm fluoroscopes (Philips Medical Systems, Best, the Netherlands), arbitrarily named the "blue" and "green" systems. The fluoroscopes' digital cameras were replaced with high speed digital video cameras (Phantom v5.2, Vision Research, Wayne NJ) capable of a 1000 Hz framerate (997 μs electronic shutter speed) at an 1152 × 896 pixel resolution. The digital cameras are configured using a mid-level workstation running Windows 7 (Microsoft, Redmond, WA) and are triggered and synchronized by a control signal sent by a microcontroller (Arduino, Ivrea, Italy). This microcontroller also activates the two fluoroscopes, which operate in continuous mode during the acquisition. The performance of the hardware has been previously described [36].

2.2. Session setup

Prior to a session of testing, the X-ray sources are positioned anterior and superior and to the medial and lateral sides of where a subject's foot would be positioned during stance gait. During manual manipulation of cadaveric feet in pilot testing, these perspectives minimized bone overlap. The image intensifier planes were positioned perpendicular to and centered on the X-ray beam. Images were captured of: (a) a *distortion correction plate* which is affixed to the image intensifiers. This rigid aluminum plate has a machined grid of 3 mm holes spaced 15 mm apart, with a unique pattern of 5 mm holes present to define the plate orientation (Fig. 1). (b) *A localizer/calibration block* that is made of a stable radiolucent polymer (R1/HG3000, GoldenWest Mfg., Inc.; Cedar Ridge, CA). The localizer block has 15 metal beads of varying diameters permanently seated within it at known locations to form a unique 3D pattern (Fig. 2). True bead centroids and diameters were determined using a coordinate measuring machine (CMM) with listed accuracy of 0.007 mm (Global Performance Model, Hexagon Metrology; North Kingstown, RI). More details are provided in the pre-processing section, and in the prior publication [36].

2.3. Validation trials

Two calcanei, tali, and first metatarsals were harvested from cadaveric donors (three females aged: 72, 80, and 82 years old weighing 53, 73, and 63 kg, respectively). The bones were embedded in foam blocks (Fig. 3, top). These blocks are rigid to prevent movement of the embedded bones, and are of low radiodensity to prevent image artifact; additionally, a plastic "wand" affixed to the block served as an attachment point for validation trials. Tantalum beads (1.6 mm diameter) were implanted in four corners of each foam block and secured with superglue. The foam thus rigidly joins the beads and the bones, but also separates the beads from the surface of the bone to reduce artifacts which occur when implanted beads in or on cortical bone are CT scanned. Validation trials were performed for each of the six bones under two conditions: translation (along the length of the walkway in the approximate AP direction) and rotation (in the transverse plane approximating internal and external rotation) of the embedded-bone foam blocks.

Each block was individually affixed to a linear stage via the wand to a 1-μm stepper-motor (ROB-09238, SparkFun Electronics, Niwok CO) with attached micrometer (Fig. 3, bottom). The choice of increment size for translational and rotational validation was based on a pilot live gait data set (not reported here). A frame-by-frame comparison of the calcaneus in this data set yielded a peak translational velocity (measured from the bone centroid) of 1.9 m/s (this occurred during toe off, after the calcaneus has left the ground); and a peak rotational velocity of 419°/s (during