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Technical note

Marker-based validation of a biplane fluoroscopy system for quantifying foot kinematics

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ABSTRACT

Introduction: Radiostereometric analysis has demonstrated its capacity to track precise motion of the bones within a subject during motion. Existing devices for imaging the body in two planes are often custom built systems; we present here the design and marker-based validation of a system that has been optimized to image the foot during gait.
Methods: Mechanical modifications were made to paired BV Pulsera C-arms (Philips Medical Systems) to allow unfettered gait through the imaging area. Image quality improvements were obtained with high speed cameras and the correction of image distorting artifacts. To assess the system's accuracy, we placed beads at known locations throughout the imaging field, and used post processing software to calculate their apparent locations.
Results: Distortion correction reduced overall RMS error from 6.56 mm to 0.17 mm. When tracking beads in static images a translational accuracy of 0.094 ± 0.081 mm and rotational accuracy of $0.083 \pm 0.068^{\circ}$ was determined. In dynamic trials simulating speeds seen during walking, accuracy was 0.126 ± 0.122 mm.
Discussion: The accuracies and precisions found are within the reported ranges from other such systems. With the completion of marker-based validation, we look to model-based validation of the foot during gait.

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1. Introduction

Studying joint kinematics in vivo allows us to quantitatively describe how the load bearing structures within our body move and function during normal use. The ability to quantify bone motions with very small errors, which is necessary to detect subtle but biomechanically significant phenomena, is greatly enhanced by improving the accuracy and precision of the primary measurements. At the same time, due to the importance of subtle motions, it is critical to ensure that the measurement process adds minimal error to joint kinematics.

Optical marker tracking has long been a standard research tool due to its non-invasive nature, flexibility in marker placement, and easy availability of hardware and software. The main limitation of optical systems is that they do not measure bone motion directly. Instead, bone position is estimated from skin-mounted markers, producing an error termed the skin tissue artifact (STA). The STA has been measured by comparing optical marker estimates of motion to direct X-ray imaging of bones in living subjects, finding single marker STA-related errors up to 4.3 mm [1] and grouped marker (cluster) errors ranging from 6.46 to 16.72 mm [2]. STAs of 3–7 mm have been found when compared to bone mounted markers in dynamic cadaver trials [3]. To further complicate the issue, STA varies by marker location, in a unique and unpredictable manner; this was explored in living subjects [4]. X-ray based radiostereometric analysis (RSA) is potentially well suited to measure bone position without STA, thus increasing kinematic accuracy. Additionally, systems using RSA do not burden the subjects with the physical attachment of measurement hardware, which may allow for a more natural gait.

Fluoroscopic systems designed for the precise capture of bone kinematics, unlike optical systems, are not at present commercially available, requiring the creation of the instrumentation in-house.

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As these devices are expected to reliably quantify motion on the sub-millimeter scale, several types of system validation are necessary to evaluate their performance. Such validation typically includes: determining the resolution of the hardware imaging chain, evaluating how the hardware and software reduce or eliminate various distortions that are inherent in such systems, and measuring static and dynamic accuracies and precisions based on precisely known positions and motions. Exhaustive validation of these systems along these lines has been previously reported [5–8].

The hardware design, software filtering and pre-processing, and marker-based validation of a new fluoroscopic biplanar system are described in the work presented here. The first and main use for this device will be imaging of the foot, though its flexibility allows for the imaging of any joint of the body.

2. Methods

2.1. X-ray generation

The overall design of our system was modified from a previously developed biplane fluoroscopic system [8,9]. Imaging is performed with a pair of Phillips BV Pulsera fluoroscopic C-arm systems (Philips Medical Systems; Best, The Netherlands). The Pulseras were structurally modified by mechanically disarticulating the X-ray generators and image intensifiers from the "C" of the C-arm, while leaving all the electrical connections in their original configuration. The detached X-ray generator and image intensifier units were then separately mounted on custom designed 5-degree-of-freedom mobile stands, built using 80/20 aluminum framing (80/20 Inc.; Columbia City, IN), that were created specifically to support and balance these components. This mounting methodology requires manual alignment of the X-ray generators and image intensifiers, but it also makes it possible to move these components independently of each other, while allowing the subjects to walk unfettered.

To reduce unnecessary X-ray exposure, the Pulseras' exposure control boards were replaced with synchronized boards, custom built by Philips, that allow a single button press to energize both systems within 80 μs. As a reference point, the estimated full body equivalent dose for a subject performing multiple (30) walking trials through the imaging area is 8 mrem.

2.2. Walkway and imaging area

A custom, hand railed walkway (1 m wide × 6.5 m long) was built using an 80/20 aluminum framework (Fig. 1). The center panel or "imaging area" panel was constructed of a radiolucent composite of thin carbon fiber panels laminated to structural foam (Accuray, ACP Composites Inc.; Livermore, CA). The aluminum frame was designed so that the central imaging panel can be removed for system alignment and distortion correction, as described below. The portion of the frame around the imaging panel allows the mounted image intensifier stands to be rolled underneath to facilitate imaging through the surface of the walkway, making it possible to image obliquely and with significant superior–inferior orientation through the foot.

2.3. Imaging and data collection

The image intensifiers have an approximate 30 cm diameter, with an active area of approximately 27 cm in diameter. The CCD camera that comes with the BV Pulsera has a maximum sampling frequency of 30 Hz, which can be too slow to image all foot bone kinematics during gait without significant loss of information. To improve image acquisition speed, the CCD cameras were replaced with high speed CMOS cameras (Phantom v5.2, Vision Research Inc; Wayne, NJ) that are capable of 1000 Hz grayscale image capture with a 997 μs shutter speed (duration of sensor exposure). The cameras have a resolution of 1152 × 896 pixels, and can collect ~3 Gb of data (approximately 3 s of data capture at maximum frame rate). The cameras are external to the Phillips imaging chain, and are connected to a separate laboratory computer for data storage. The cameras, and the modified dual fluoroscopes, are triggered simultaneously by a single custom built switch box.

2.4. Alignment

Each time the X-ray generator is moved, it must be aligned such that its beam is both normal to and centered on the image intensifier. Misalignment reduces image contrast and brightness, wastes X-ray exposure and can also create a non-uniform smear in the image and reduce field of view. Alignment is achieved using a guide laser built into each X-ray generator to determine if the source and image intensifier are parallel and centered.

2.5. Distortion correction, bias correction, and localization

These pre-processing algorithms were custom written in Matlab (MathWorks, Natick, MA).

2.5.1. Distortion correction

There are two major sources of distortion in the data collection chain [10], namely pincushion distortion and magnetic lens distortion. To correct for these distortions, a round aluminum plate with laser cut 3 mm diameter holes equally spaced 1.5 cm apart is rigidly fixed to the input side of the image intensifier. A few specifically placed 5 mm holes allow for the automatic detection of the orientation and center of the image intensifier. After being imaged on each fluoroscope, the known size and pattern of the centroids of the holes in the calibration plate are used as control points. The imaged locations of these control points are then used with a thin plate spline algorithm (approximation method) to generate a correction map, which allows us to spatially re-map (correct) every distorted image [11]. The root mean square (RMS) error between the image points and the control points was calculated before and after correction.

2.5.2. Bias correction

The final image quality issue deals with intensity bias which arises from the variation of pixel saturation across an image [10]. As a result, the images captured show a non-uniform brightness which varies in space. The effect of this is that both the beads (and later bones) will vary in their intensity and contrast depending on their location in the image field. Accounting for this bias would increase the consistency of the image characteristics (i.e., contrast and raw intensity values). To correct for this, at every exposure setting (kV and mA) used for data collection or for post processing, a shot is taken with nothing between the X-ray sources and image intensifiers to give an estimate of the intensity distribution in space. The maximum single pixel intensity of this blank image is then subtracted from the whole image to yield a bias map for each image intensifier. This bias map is then applied to every image obtained to normalize the intensity across the image. While considerable image uniformity improvement results from these corrections, in practice they do not create a perfectly uniform background, as the presence and movement of objects in the beam further alters intensity.

2.5.3. Localization

For accurate 3D motion correlation from stereoscopic imaging, the location and orientation of the image intensifiers relative to one another must be precisely known. To obtain this information we created a 3D calibration block from dimensionally