Upper Extremity Kinetics During Lofstrand Crutch Assisted Gait

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Introduction: The number of persons with Spinal Cord Injury (SCI) has increased significantly in the past 20 years. With improved treatment and care, individuals with neurologically incomplete SCI can achieve functional ambulation with proper bracing and walking aids. The use of assistive devices places added loads upon the upper extremities, particularly the glenohumeral joint, which was not designed for weight bearing. Over time, the high weight bearing loads on the shoulder are not well tolerated. Consequently, individuals with incomplete SCI are developing disabling shoulder joint pain at higher rates than normal, similar to that seen in individuals with complete injuries.

Forces and moments applied to the handles of walkers, crutches and canes during device assisted gait has been reported.[1,2] Liggins, et al. instrumented a Lofstrand crutch with a six component load cell and three reflective markers (Vicon) in order to model the crutch kinematics.[3] Extending the model to include the human upper extremity and crutch system in order to determine the forces and moments at the joints during Lofstrand crutch assisted gait has not been reported.

The objective of this work was to develop a kinematic and kinetic model of a Lofstrand crutch and upper extremity and to instrument the crutch with transducers that measure the forces and moments applied to the arm by the crutch.

Clinical Significance: To prevent further loss of functional independence for individuals with incomplete SCI, it is imperative to find ways to reduce the strain and joint deterioration that may occur with crutch use.

Methodology: <u>Instrumentation:</u> A pair of Lofstrand crutches were instrumented with 6-component load cells (Bertec) placed just below the handles. Foil strain gauges (Micro Measurements) were mounted just above the handles, to measure the cuff moments. A sixteen-channel EMG system (Motion Lab Systems) was modified by the manufacturer to extend the lower end of the bandwidth to DC. This system was used to multiplex the crutch data over a thin co-axial cable, for data acquisition. Footswitch data were telemetered from compression closing footswitches (B & L Engineering) to define the gait cycle. Reflective markers were placed at the tip of the crutch handles and at two locations on the crutch shafts (Fig. 1). Reflective markers were placed on each upper extremity and the trunk to define the 3D motion of upper extremity body segments. Crutch data were sampled and digitized at 2500 Hz on a DEC PDP 11/73 computer. Vicon motion data were recorded at a 50 Hz frame rate on a second DEC PDP 11 computer. Motion and force data were synchronized and processed on a DEC VAX computer.

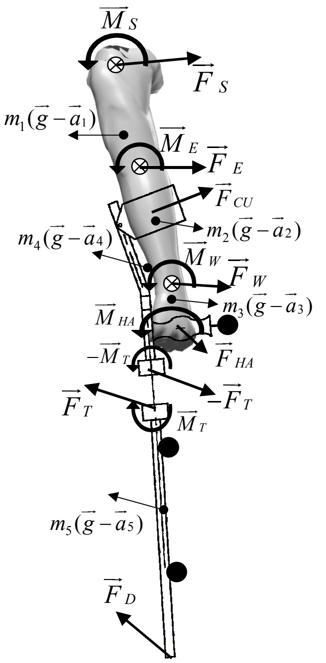

Figure 1.

Model: An inverse dynamics model (Fig. 1) of the human upper extremity + crutch system was formulated to determine the upper extremity joint loads during crutch-assisted gait. Considering the shoulder loads ($\vec{F}_S$ and $\vec{M}_S$) for instance:

$$\vec{F}_S = \vec{F}_D + \sum_{i=1}^{5} m_i(\vec{g}-\vec{a}_i), \quad \vec{M}_S = \vec{r}_{SD}\times\vec{F}_D + \sum_{i=1}^{5}\vec{r}_i\times\left[m_i(\vec{g}-\vec{a}_i)\right] - \sum_{i=1}^{5}\vec{M}_{I_i}$$

The crutch distal force $\vec{F}_D$ was determined from the transducer output $\vec{F}_T$: $\vec{F}_D = -\left(\vec{F}_T + m_5\left[\vec{g}-\vec{a}_5\right]\right)$, where m_i is ith segment mass, $\vec{g}$ and $\vec{a}_i$ are gravitational and inertial acceleration, respectively, $\vec{r}_{SD}$ and $\vec{r}_i$ are vectors from the shoulder joint center to crutch tip and ith segment mass center, respectively, and $\vec{M}_{I_i}$ is the inertial moment due to the ith segment. Similarly, determination of the individual segment (e.g. $\vec{F}_{HA}$, $\vec{M}_{HA}$) and joint (e.g. $\vec{F}_E$, $\vec{M}_E$) loads incorporated the above equations plus the measured cuff force $\vec{F}_{CU}$ and the derived transducer moment $\vec{M}_T$.

Results: Glenohumeral joint motion (Fig. 2) and force (Fig. 3) in trunk coordinates are shown for a typical SCI patient. The vertical dashed lines represent the transition from stance to swing of the contralateral lower extremity. Peak superiorly directed force of 80 N occurred at about 20% of the cycle.

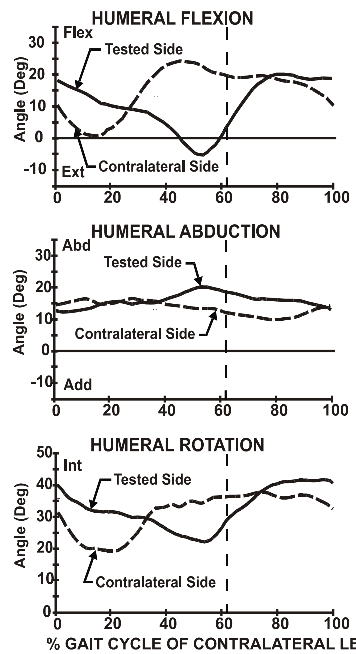

Figure 2.

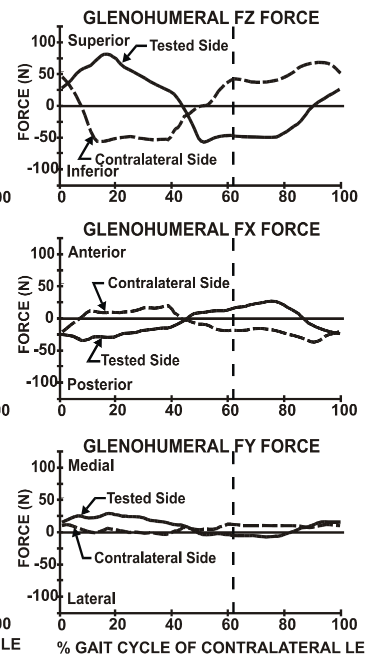

Figure 3.

Discussion: This model allows us to investigate the upper extremity weight bearing forces during ambulation with assistive devices. Using footswitches to define the gait cycle allows clinicians to determine the relationship of shoulder joint forces to the lower extremity weight bearing events.

References:

1. Murray MP et al. Am J Phys Med., 48:1-13, 1969.
2. Melis EH et al. Spinal Cord., 37:430-439, 1999.
3. Liggins AB et al. Gait & Posture, 13:295-296, 2001(Abstract).

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