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Analysis of terrain effects on the interfacial force distribution at the hand and forearm during crutch gait

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ABSTRACT
Forces transferred to the upper body during crutch use can lead to both short-term and long-term injuries, including joint pain, crutch palsy, and over-use injuries. While this force transmission has been studied in controlled laboratory settings, it is unclear how these forces are affected by irregular terrains commonly encountered during community ambulation. The purpose of this study was to determine the effects of walking speed and uneven terrain on the load magnitude, distribution, and rate of loading at the human-crutch contact surfaces. Our results show that the rates of loading were significantly increased with higher walking speeds and while negotiating certain irregular terrains, despite there being no apparent effect on the peak force transmission, suggesting load rate may be a more appropriate metric for assessing terrain effects on crutch gait. Furthermore, irrespective of the type of terrain and walking condition, the largest compressive forces were found to reside in the carpal-tunnel region of the hand, and may therefore be a primary contributor to carpal-tunnel injury.

Introduction
Crutches are commonly prescribed to enable functional mobility for individuals with walking impairments, with over 566,000 crutch users in the United States alone (Kaye, Kang, & LaPlante, 2000). These users generally fit into two categories: short-term users who rely on crutches to lessen the weight placed on an injured limb, and long-term users who require the stability provided by crutches because of pathologies such as spina bifida, lower limb amputation, or cerebral palsy. Over 40% of crutch users have reported inability to perform many everyday activities due to their device, such as ascending and descending stairs, using public transportation, and navigating the outside environment (Kaye et al., 2000). Furthermore, prolonged crutch-use has been linked to additional short-term and long-term complications, including joint pain, crutch palsy, aneurysms, and repetitive use injuries (Ackermann & Taissun, 2012; Opila, Nicol, & Paul, 1987; Segura & Piazza, 2007). One such injury, carpal-tunnel syndrome, which causes pain or numbness in the hand and wrist due to compression of the carpal-tunnel nerve, is particularly common (Fischer, Thompson, & Harrison, 2014; Sala, Leva, Kummer, & Grant, 1998). This compression occurs most often due to an increase in force applied in the carpal-tunnel region of the hand, near the base of the thumb.

The aforementioned complications, especially carpal tunnel syndrome, have been primarily attributed to increased forces transmitted to the upper extremities during crutching, and a number of studies have looked to determine the effects of loading on the joints and muscles of the upper limbs during ambulation (Bhagchandani, 2009; Goh, Toh, & Bose, 1986; Opila et al., 1987; Requejo et al., 2005; Segura & Piazza, 2007). For example, Goh et al. used force transducers placed on both the proximal and distal ends of an axillary (under-arm) crutch to determine that the palm of the hand supports 44.4% of bodyweight at peak force during the gait cycle (1986). Similarly, Bhagchandani (2009) used load cells placed just above and below the handles of a Lofstrand (elbow) crutch to isolate the forces transmitted through the elbow and hand, respectively. A study by Sala et al. (1998) placed pressure sensors to directly measure contact force between the user’s palm and crutch handle. By using an array of six sensors distributed across the hand, they were able to determine how force is distributed across the hand for two different handle designs.

Previous investigations have almost exclusively been performed in laboratory settings on uniform, flat surfaces. However, forces measured during these conditions may not accurately reflect those present in real-world conditions with variable environments. Previous research has shown that traversing irregular terrain increases gait variability, ground reaction force, and upper-body instability during unassisted walking (Andrysek, Klejman, & Kooy, 2014; Chou & Draganich, 1997; Marigold & Patla, 2008; Thies, Richardson, & Ashton-Miller, 2005) which all have important clinical implications on the in-community mobility of assistive-device users; however, little has been done exploring the effects of these environments on crutch use. Therefore, a more complete analysis of upper-limb
force transmission would seek to replicate everyday conditions, such as variable walking speed and uneven terrain.

The purpose of this study was to determine if walking speed or uneven terrain significantly affect the interfacial load magnitude, interfacial load distribution, ground reaction force, and rate of force rise that could potentially affect the onset of carpal tunnel syndrome during ambulation using forearm crutches (Goh et al., 1986; Sala et al., 1998; Segura & Piazza, 2007). Pressure sensors on the crutch handle, and a load cell incorporated into the crutch shaft were used to record the interfacial forces and load rate, respectively. We hypothesized that environmental changes would increase the interfacial forces and force rate, and that these forces would be largest in the carpal-tunnel region of the hand.

Methods

Ten able-bodied individuals (eight males and two females), aged 24 ± 2.5 years with mass 77.72 ± 15.27 kg (171.3 ± 33.66 lb), volunteered as test subjects. Participants were free from cognitive impairments and did not currently require the use of a mobility aid. All procedures were approved by the Research Ethics Boards of Holland Bloorview Kids Rehabilitation Hospital and the University of Toronto.

Instrumentation

A pair of forearm crutches (Sidestix Ventures Inc, Sechelt, Canada) was used for all trials, with one crutch modified to include a 6-axis load cell (MC1.75, Advance Manufacturing Technologies, Crowthorne, UK) in the shaft slightly below the handle (Figure 1). This location was chosen to reduce the weight at the crutch tip to almost nearly approximate normal gait. The crutches had an optional spring damping component that was disabled during all experiments to make a rigid crutch. The load cell was sampled at 100 Hz, which was recorded by data acquisition hardware (CRONOS-PL, imc DataWorks, Novi, MI, USA) housed in a backpack worn by the researcher walking behind so as not to pace the participant, tethered by a cord. Data were then transmitted wirelessly to a laptop computer for processing using a local area wireless network.

In addition to the load cell, flexible piezoelectric force sensors (FlexiForce A201, Tekscan Inc, South Boston, MA, USA) were mounted on the crutch handle, as well as on the contact surface of the forearm support. Each sensor was 9.53 mm (0.375 in.) in diameter with 0.208 mm (0.008 in.) thickness, and measured only the forces normal to the sensor surface. Four sensors were placed on the forearm support (Figure 2c), and six sensors were arranged on the upper surface of the handle (Figure 2b) to measure the contact pressure at discrete portions of the palm, as shown in Figure 2a. Additionally, one sensor was included on the bottom of the handle to measure grip force of the fingers as they oppose the palmar forces on the opposite side (Figure 2d). The FlexiForce recording system was mounted directly to the crutch itself, and sampled at 50 Hz. Due to the FlexiForce sensor sensitivity to bending and soft materials (Brimacombe, Wilson, Hodgson, Ho, & Anglin, 2009; Ferguson-Pell, Hagisawa, & Bain, 2000), each sensor was calibrated before each test with known weights to establish the voltage-force relationship.

Crutch height was adjusted to suit each participant as per the manufacturer’s guidelines. Each participant was subjected to six different walking scenarios. Three were on flat ground in a laboratory environment, and three outdoors: over rocky terrain, up a ramp, and down a ramp. All trials were completed over a distance of 10 m. Prior to testing, participants were instructed on how to use the crutches by the research team. A swing-through gait was selected because it ensures that 100% of the participant’s bodyweight is supported by the crutch at some point during ambulation. The instrumented crutch was placed under the right arm of all participants. After becoming comfortable with the use of the crutches, which was assessed via observation and feedback from the user, the participants were asked to walk at a self-selected “normal” rate for the 10-m span, using the swing-through gait. After three trials, the participants were then asked to walk as fast as comfortable for three trials, and then as slowly as comfortable for three trials. The time elapsed for each 10-m trial was recorded and used to calculate average velocity at each self-selected speed.
Participants were then taken outdoors to perform walking trials under three conditions. The conditions included uneven ground with scattered rocks with an average diameter of approximately 10 cm (3.94 in.), and walking up and down an outdoor access ramp with an 8° slope (13% grade) and 10 m length. The order in which the conditions were tested was randomized. The participants were allowed to choose their own walking speed, and three trials were recorded for each condition.

Data processing
LabVIEW (National Instruments, Austin, TX, USA) was used to process the FlexiForce data, and imc-Devices (imc DataWorks, LLC, Novi, MI, USA) to process the load cell data. Ground reaction force was measured directly from the load cell force component corresponding to the long-axis of the crutch. Ground reaction force measured by the load cell was numerically differentiated with respect to time to calculate the load rate in the crutch. The interfacial force and axial crutch force with the highest magnitudes at each walking step were selected for analysis, resulting in between 6–9 data points for each participant in every environment. To enable more direct comparison to previous results in the literature, force measurements were recorded in Newtons and then normalized to patients’ bodyweight to determine the percent bodyweight of each load. Similarly, the average pressure across each sensor was computed by dividing the force measurements by the sensor area (77.33 mm²).

Statistics
All statistical calculations were performed using the SPSS 21 (IBM, Armonk, NY, USA). Two-way analysis-of-variance (ANOVA) with Tukey post-hoc tests were performed on the FlexiForce data to identify interfacial force differences caused by both environmental effects and sensor location. One-way ANOVA was used to identify environmental effects on the ground reaction force, rate of force rise, and self-selected walking speed. For all tests, repeated measures techniques were used to account for multiple trials per subject, and significance levels were set at $\alpha = 0.05$.

Results
Walking speed
The average self-selected walking speed for each of the six test conditions is shown in Figure 3a. As expected, significant differences were seen among all three walking speeds ($p < 0.001$). The self-selected walking speed for the “Rocky” condition was significantly different from all but the “Slow” speed condition. Similarly, the “Up” and “Down” conditions were significantly different from both the “Slow” and “Fast” speeds, but did not differ from each other or the “Normal” condition. Therefore, for this investigation it is most appropriate to use the “Slow” condition as a baseline for comparing force and force rate measurements of the “Rocky” trials, and the “Normal” condition as a baseline for the “Up” and “Down” trials.
As expected, both walking speed and terrain significantly affected the maximum rate of force increase in the crutch shaft (Figure 3b). The “Fast” walking speed exhibited significantly higher load rates ($935.2 \pm 469.7 \text{ % \, s}^{-1}$) than both the “Slow” ($342.9 \pm 151.0 \text{ % \, s}^{-1}$) and “Normal” ($419.8 \pm 215.7 \text{ % \, s}^{-1}$) conditions with significance $p < 0.001$, but was not significantly different from the “Rocky” ($p = 0.117$) or “Down” ($p = 0.088$) conditions. When accounting for walking speed, comparison of the “Rocky” and “Slow” conditions reveal significantly larger load rates for the “Rocky” group ($p = 0.027$). However, neither the “Up” nor “Down” conditions evidence any significant differences from the “Normal” speed trials ($p > 0.273$). Therefore, we see that among the different terrains explored, the uneven “Rocky” environment had the most pronounced effect on rate of force increase.

**Maximum interfacial force**

Flexiforce signals for the hand and forearm, respectively, were summed to estimate total force transferred from the crutch to the upper limb. Figure 4 illustrates the mean and standard deviation of the axial crutch load, total hand force, and total forearm force for one gait cycle. During testing, the location of Sensor 2 on the handle (Figures 2b and 2d) resulted in incomplete or inconsistent contact with the hand for several participants. This incomplete contact was confirmed by the resulting sensor measurements, which exhibited interfacial forces between 3–6% bodyweight for a few participants, and near-zero values for the remainder. Due to this high inter-subject variation, the data from the hand Sensor 2 was not considered in the subsequent analyses. Similarly, Sensor 1 and Sensor 2 on the forearm exhibited highly fluctuating signals, and were not included in the statistical analysis. The total force carried in the hand

**Axial load rate**

As expected, both walking speed and terrain significantly affected the maximum rate of force increase in the crutch shaft (Figure 3b). The “Fast” walking speed exhibited significantly higher load rates ($935.2 \pm 469.7 \text{ % \, s}^{-1}$) than both the “Slow” ($342.9 \pm 151.0 \text{ % \, s}^{-1}$) and “Normal” ($419.8 \pm 215.7 \text{ % \, s}^{-1}$) conditions with significance $p < 0.001$, but was not significantly different from the “Rocky” ($p = 0.117$) or “Down” ($p = 0.088$) conditions. When accounting for walking speed, comparison of the “Rocky” and “Slow” conditions reveal significantly larger load rates for the “Rocky” group ($p = 0.027$). However, neither the “Up” nor “Down” conditions evidence any significant differences from the “Normal” speed trials ($p > 0.273$). Therefore, we see that among the different terrains explored, the uneven “Rocky” environment had the most pronounced effect on rate of force increase.

**Figure 3.** (a) The average self-selected walking speed across all 10 subjects is shown for each condition. The upper and lower box limits indicate the 75th and 25th quartiles, respectively, and the “x” identify individual data points. Data extending beyond the dashed lines were considered outliers and not considered in the analysis. The “**” indicates significant differences ($p < 0.001$) from all “no-” groups, and the “†” indicates significant differences ($p < 0.001$) from all “non-” groups. Walking speeds for the “Rocky” condition were closest to those selected for the “Slow” trials, while both “Up” and “Down” conditions were most similar to the “Normal” trials. (b) Peak load rate measured in the crutch shaft for each speed and environment. Bars with alike markers (”,+,”) are not significantly different from one another, but are different non-alike groups ($p < 0.001$). The “Rocky” and “Fast” environments exhibited the highest force rates, the mean load rates for the rocky terrain exhibiting a significant increase from the “Slow” condition, indicating that walking speed alone does not account for the increase.

**Figure 4.** Crutching forces as a function of gait cycle. The mean (lines) and standard deviation (shaded area) of all nine participants are shown for the crutch axial force (-), summed hand forces (--), and summed forearm forces (-–).
and forearm is shown in Figure 4 as a percentage of bodyweight. Due to the relatively large spread of the interfacial force data, the ANOVA showed no significant differences in total force carried by the hand for any of the environments ($p = 0.159$). Furthermore, the forearm carried significantly less weight than the hands in every environment ($p < 0.001$). As expected, based on the load cell data a single crutch carries approximately 50% of the bodyweight at peak loading in a swing-through gait (Figure 4). The summed hand loads and crutch force exhibited similar trends during the gait cycles, with the difference between these metrics being at least partially attributed to any user-crutch contact that may have occurred between sensors.

**Interfacial force distribution**

Figure 5 shows the portion of the participants’ bodyweight carried by each of the included sensors for all six environments. Sensor 4, which was located almost directly above the carpal-tunnel, carried the majority of the hand load, roughly 10% of total bodyweight, for each environment. The environment did not exhibit any significant effects on the proportion of load carried by each sensor ($p = 0.913$), with the relative portion of the total bodyweight carried in each region of the hand remaining fairly constant across conditions.

**Grip force**

Grip strength (Figure 6) remained nearly constant across trials, with the only significant difference being an increase in gripping force for “Fast” ($2.37 \pm 0.67\% \cdot s^{-1}$) walking speeds relative to “Slow” ($1.15 \pm 0.52\% \cdot s^{-1}, p = 0.002$). As with load rate and hand load, the high degree of variation in the data masks trends apparent in the figures, which may suggest an increase in grip force in “Rocky” condition relative to its speed-matched “Slow” condition.

**Discussion**

Large forces transmitted to the upper body during crutch use can cause a number of complications, including carpal-tunnel syndrome. While this force transmission has been studied in laboratory settings, to the best of the authors’ knowledge, little has been done to explore how this transmission is affected by different walking speeds and terrains, conditions that are commonly associated with crutch use. Hence, the principal
contributions of this study include the assessment of biomechanically and clinically relevant aspects of crutch loading under previously unexplored mobility conditions. In summary, experiments were conducted to assess interfacial forces at the hand and forearm, ground reaction force, rate of force rise, and walking speed and data were examined to determine the effects of walking speed and uneven terrain on the resulting force and magnitudes and distributions.

The high force values reported by Sensor 4 indicate that a high percentage (up to 15%, as seen in Figure 5) of the participants’ bodyweight was transmitted directly through the carpal-tunnel region of the hand, which agrees well with the values reported by of Sala et al. (1998), who saw between 14–20% of the hand load in this region. The distribution of load across the hand and forearm was fairly consistent across the different environments, indicating that participants’ weight distribution was not affected by speed or terrain. The high compressive forces in the carpal-tunnel region, regardless of environment, likely appreciably contribute toward the onset of carpal-tunnel syndrome. Therefore, diverting these compressive forces from the carpal-tunnel region to other areas of the hand or forearm may be potential mechanisms by which to reduce the likelihood of overuse injury. For example, improvements to the handle design to better contour to the hand and increase the surface area near the carpal tunnel region to help minimize the local pressures and redistribute the load.

The total pressures in the hand were much higher than the290 kPa (42 psi) pressures reported by Sala et al. (1998). These increased pressures can most likely be attributed to the additional weight carried by the upper body during swing-through gait compared to the three-point gait employed by Sala et al. In addition to this increase, our results show markedly different load distributions than reported in their study, which can most likely be attributed to the dissimilar handle geometries between the two studies. Of the two handle designs explored by Sala et al., our results showed load distributions more similar to their wider handle design. The total forearm pressures seen in our study were similar to the combined pressures of the medial and intermediate regions reported by Fischer, Nüesch, et al. (84 – 208 kPa; 2014), with these regions corresponding to the approximate location of our forearm sensors.

Contrary to our hypothesis, neither walking speed nor environment had a significant effect on either the interfacial load magnitude or distribution. While this lack of statistical significance may in part be explained by the pronounced variation in the FlexiForce data, these results may also suggest that load rate rather than force magnitude alone should be used as a metric for evaluating environmental effects on crutch use. The increased load rate between the “Slow” and “Rocky” trials is particularly interesting since the average walking speed for these two conditions were not significantly different, indicating the change in load rate is due to the difference in terrain. Some of this increased load rate can be attributed to local fluctuations in the ground reaction force shown in Figure 4, which may be caused by shaking or the crutch tip slipping over the rocky surfaces. These sudden movements introduce repeated small impacts that could serve to destabilize the crutch user and lead to a perceived sense of imbalance and discomfort. The decreased walking speed, relative to normal, seen in the rocky trials supports this possibility. The participants were walking slower and being more careful over the rocky terrain, but still evidenced higher load rates, signifying that the terrain and not the walking speed may have caused the increase. In addition to affecting walking stability, the repeated impacts suggested by the higher load rates can lead to overuse injuries in the upper body extremities (Feldman, Vujic, McKay, Callcott, & Uflacker, 2010).

Load rate has previously been explored by Segura and Piazza (2007) as an indicator of jarring or quickly changing impact forces. Our results exhibit similar values and trends, suggesting load rates tending to be smaller for the more stable conditions (“Slow” and “Normal”) and higher for the more extreme speeds and terrains. In their study, Segura and Piazza also explored the use of spring-loaded crutches as shock-absorbers to minimize impact forces present during walking. They saw that spring-loaded crutches worked to decrease the maximum force rate, but at the cost of decreased walking speed and reported decrease in participants’ feeling of stability. However, this perceived instability may have been due to inexperience with the spring-loaded crutches. The decreased load rate afforded by the spring-loaded crutches shows that they may be particularly beneficial in absorbing some of the load fluctuations caused by the uneven terrain, but the specific effects of environment on the performance of spring-loaded crutches have yet to be examined.

None of the participants in this study had prior experience using crutches, which biases the potential application of these findings toward short-term users. Specifically, experienced crutch users may adapt their crutching style to alleviate contact pain, resulting in techniques that differ from the manufacturer’s suggested technique used in this study. Given the relatively small forearm forces presented, this possibility is supported by the prevalence of overuse injuries in the forearm associated with long-term elbow crutch use (Ginanneschi, Filippou, Milani, Biasella, & Rossi, 2009; Malkan, 1992; Venkatanarasimha, Kamath, Kambouroglou, & Ostlere, 2009). As such, a better understanding of the balance between hand and forearm loads for long-term and short-term crutch users needs to be reached. These differences may help identify any coping strategies used to minimize crutching related pain and injury, which could both inform future changes to crutch design, and possibly be taught to less experienced users as preventative measures. This study presents a first approximation of the interfacial force distributions exhibited with short-term crutch users, and future studies will aim to evaluate and quantify force transmission in more experienced crutch users, which can help inform new and safer crutch design.

It should be noted that the total force values (Figure 4) reported in this study do not account for the entire load carried by the hand and forearm, leaving, on average, 20% of the bodyweight load unaccounted for. This missing force is transmitted from the crutch to the body through other interfacial surfaces, such as the area between sensors, particularly the region around the hand Sensor 2 and forearm Sensor 1 and Sensor 2 where contact between the participant and sensor was inconsistent.
Additional measurement error stems from the moderate-to-good reliability of the FlexiForce sensors. Although the manufacturer’s documents for the sensors specify a ±2.5% reliability, previous studies have reported errors up to 20% (Brimacombe et al., 2009; Ferguson-Pell et al., 2000; Zammit, Menz, & Munteanu, 2010), with a similar degree of variability evidenced in our data. The variation exhibited in this study is partially due to the sensors being subjected to large deformations that can adversely affect the accuracy of the output signal (Ferguson-Pell et al., 2000). Care was taken to calibrate each sensor before every set of trials, but deformation effects caused by surface curvature likely still introduced some error into the signals during the experiments, resulting in increased variability. Improvements in transducer performance will be essential for future research aiming to quantify and understand the forces at body and assistive device interfaces.

Conclusion

Neither walking speed nor the introduction of uneven terrain exhibited significant effects on interfacial force magnitude or distribution during crutching, however effects on load rates were found. In all environments and for all speeds, the maximum force transmitted from the crutch to the upper body occurred through the carpal-tunnel region of the hand, which may contribute to the onset of carpal-tunnel syndrome. Furthermore, the lack of difference in force magnitude and distribution between different walking speeds and environments, combined with the differences seen in load rate, indicate that the latter metric may be more important in evaluating environmental effects on crutch gait. The data presented in this study were collected from healthy, inexperienced crutch users, and may not be relevant for users with mobility impairments. Additional work is needed to explore how these load distributions and load rates vary in more experienced crutch users, which can ultimately be used to inform improved crutch design and minimize crutch-induced injury.

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