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finger tremor amplitude results from the mechanical influence of cardiac activity. It is well known that tremor and fluctuations in force output share neuromuscular underpinnings (e.g., motor unit firing behavior) (Marsden, 1984, Moritz et al., 2005). Therefore it is possible that CBI contributes in part to fluctuations in force output.

In one of the few recent investigations of CBI influence on force production, Gallasch and Kenner (1997) suggested that there was a decrease in the cardioballistic contribution to force fluctuations during increasing force production in a grip task—but only visible assessment of waveforms was conducted. It is well known that the level of force produced is linked to the amount of force variability (Schmidt et al., 1979). Consequently, it is not clear if the contribution of CBI will be influenced by the amount of force produced.

Although an issue of recent debate (c.f. Tracy et al., 2007, Baweja et al., 2009), it is maintained that visual feedback minimizes the variability of isometric force production (Slifkin et al., 2000; Sosnoff & Newell, 2005). Previous investigations of CBI and force production have not altered visual feedback. So it is not clear whether visual feedback will influence CBI on force variability. Given the seemingly rhythmical (i.e., predictive) nature of CBI it is possible that visual feedback of force output will enable subjects to be able to correct for perturbations resulting from CBI.

Consequently, the purpose of this investigation is to examine whether there is a contribution of CBI to fluctuations in force output and whether it is mediated by force level or visual feedback. To address this research question subjects produced continuous isometric force output to a range of force levels (0.5, 1 and 2 N) with and without visual feedback while their finger tip pulse pressure was simultaneously recorded. It was predicted that there would be a significant association between cardiovascular activity and force output and that this association would be greatest at low force levels in the no vision conditions. This prediction was based on the assumption that CBI is relatively small (Gallasch & Kenner, 1997) and visual feedback would allow for corrections of any perturbation.

Methods

Subjects

A total of 11 subjects (5 males) with a mean age of 23.9 years old (SD 2.1 years) participated in the investigation. To be included in the study, participants met the following selection criteria: a) were free of any neuromuscular disorders, finger or wrist abnormalities; b) were normotensive (i.e., resting brachial blood pressure < 140/90 mmHg); c) had normal or corrected to normal vision; and d) were right hand dominant. The subjects signed informed consent forms that had been approved by the University of Illinois at Urbana-Champaign Institutional Review Board.

Apparatus

Isometric Force Recording. A schematic of the experimental set-up is presented in Figure 1. Participants were seated in a chair facing a 17-inch video monitor (DELL), with their dominant hand placed in a prone position on a table top; so that their shoulder was abducted at 45° and their elbow flexed at 90°. Orthogonal
to the index finger was an Eltran ELFS-B3 load cell (diameter 1.27 cm). The load cell measured compressive force produced by the index finger. The anterior side of the distal portion of the index finger was in constant contact with the load cell producing compressive force.

Voltage changes from the load cell were amplified by a Coulbourn (V72–25) resistive bridge strain amplifier with an excitation voltage of 10 V and an amplifier gain of 100. The analog force signal was sampled at 200 Hz by a 16-bit analog to digital converter (Biopac, Inc.). The smallest increment of change in force that could be detected was 0.002 N.

At each sampling interval the force produced was presented on the monitor. The monitor had a viewing area of 1200 horizontal pixels and 1000 vertical pixels. The monitor was approximately 50 cm from the volunteer’s eyes and 100 cm from the ground.

**Brachial Artery Blood Pressure Assessment:** Resting blood pressure was measured in a seated position using an automated oscillometric cuff (HEM-712C, Omron Corporation, Japan) before and immediately following the force production task. All brachial BP measurements were made in duplicate and the average value recorded.
**Beat-to-Beat Blood Pressure Acquisition:** With the subjects in a seated position, beat-to-beat blood pressure was recorded using finger plethysmography (Finometer, FMS, The Netherlands) with the arm maintained at heart level. The blood pressure cuff was attached to the 3rd digit of the dominant hand. Validity and reliability of this noninvasive technique has been demonstrated against invasive methods and has been shown to provide accurate measurement of blood pressure changes when compared with intra-arterial blood pressure (Imholz et al., 1991). All signals were sampled at 200 Hz and written to disk for offline analysis with the use of a BioPac system (Biopac, Inc).

**Procedures**

**Experimental Design and Instructions.** After informed consent was provided, subjects were seated in a comfortable position and allowed to rest for 5 min before resting blood pressure being recorded. The finger blood pressure cuff was then attached to the subject. Once the subjects were outfitted with the cardiovascular equipment the force control paradigm was initiated.

The participant adjusted their force output to match a target line displayed on the midline of the monitor and viewed online feedback of their performance in the form of a continuous line which corresponded to the force trajectory that moved left to right across the screen with time. The target line corresponded in independent conditions to 0.5, 1 and 2 N. Force levels were based on previous research (Sosnoff, Vallantine & Newell, 2006). Trials were performed with and without visual feedback. In the no vision condition, there was a presentation of visual feedback for the initial 7.5 s of the 30 s trial so that subjects could achieve the force target. The visual scale of the monitor was constant across conditions and resulted with a 1.5 N scale above and below the target line.

Each testing session consisted of 18 trials (3 force levels × 2 visual conditions × 3 trials). The force levels were randomly presented within a block of 6 trials. The no vision condition was performed at the end of each block of trials to insure that subjects were familiar with the force requirement. Three 30 s trials were performed at each unique force-vision condition. Adequate rest was given between trials to minimize fatigue with a minimum of 30 s between trials and 45 s between blocks. Participants were instructed to minimize the deviations between the force trajectory and the target line throughout all trials. Following the completion of the force control paradigm resting blood pressure was accessed.

**Data Analysis**

The initial 9 s and final second of force data from each trial were removed before analysis to avoid the initial force stabilization, alteration in force output with removal of vision and premature cessation of force production. All force data processing was performed using a custom written program written in MatLab v7 (The Mathworks).

**Task Performance**

To access task performance as a function of force level and visual feedback distributional statistics (i.e., within trial mean, and standard deviation (SD)) and root mean square error (RMSe) of the force data were calculated.
Cardioballistics

Central to this investigation is the question as to whether fluctuations in force output are related to cardiovascular activity (i.e., cardioballistics). To index CBIs the association between force output and beat to beat blood pressure was quantified in the frequency domain via coherence analysis. Analogous to correlation in the time domain, coherence analysis is a means of accessing a relationship between two time series in the frequency domain. It provides an estimate of how well a time series corresponds to another time series at each frequency. Coherence is a function of frequency and ranges from 0 to 1 with a stronger coupling indicated by a value closer to 1. A window size of 512 points and a sampling rate of 200 Hz resulted in a 0.39 Hz frequency bin for each power spectral estimate. In order for coherence analysis to be computed, the power spectra of the force signal (\( \chi_f \)) and the blood pressure (\( \chi_{bp} \)) signal were first determined. Subsequently, the cross-spectra of the force and blood pressure time series were determined (\( \chi_{f-bp} \)). Consequently, coherence (R) is determined by the following equation:

\[
|R_{f-bp}|^2 = \frac{\left| \chi_{f-bp} \right|^2}{\chi_f \chi_{bp}}
\]

Changes in the coherence due to force level and visual feedback were evaluated by calculating maximal coherence between 0–30 Hz and the frequency at which it occurred. 0–30 Hz bandwidth contains the vast majority of the fluctuations in force output. Significant coherence values were determined utilizing the method of Rosenberg and colleagues (1995).

Statistical Analysis

The dependent variables discussed above were each placed independently in a two-way (3 \( \times \) 2) repeated measure analysis of variance (ANOVA) with force level and visual feedback as within group factors. When relevant, Tukey’s Honestly Significant Difference (HSD) test was used to determine the specific effects contributing to the general ANOVA. All statistics were evaluated as significant when there was less than a 5% chance of making a Type I error (\( p < .05 \)), and only significant effects are reported. All statistical analyses were completed using SPSS statistical package (SPSS, Inc.).

Results

Brachial Artery Blood Pressure:

There was no change in systolic blood pressure (118 vs. 117 mmHg) from rest to post force paradigm nor was there a change in diastolic blood pressure of (70 vs. 72 mmHg) (\( p \)'s > 0.05). Overall, these findings suggest that there was minimal influence of low level isometric force production on cardiovascular function.

Task Performance:

Statistical analysis of mean force output revealed that subjects were capable of scaling their force output to the prescribed force targets with a main effect of force
level \([F(2,20) = 1239; \ p < .05, \ \eta^2 = 0.99]\). Post hoc analysis revealed that the 3 force targets were significantly different from each other (0.49 N, 0.99 N and 1.94 N, respectively). It was also found that visual feedback \([F(1,20) = 6.1; \ p < .05, \ \eta^2 = 0.38]\) influenced mean force output, with the no vision condition (1.11 N) having less force output compared with the visual feedback condition (1.17 N).

Statistical analysis of force SD revealed a main effect for force level \([F(2,20) = 33.7; \ p < .05, \ \eta^2 = 0.77]\), visual feedback \([F(1,10) = 36.5; \ p < .05, \ \eta^2 = 0.79]\) and an interaction between force level and visual feedback \([F(2,20) = 14.1; \ p < .05, \ \eta^2 = 0.59]\). The interaction between force level and visual feedback resulted from the lack of difference in force SD between the 0.5 and 1 N targets with visual feedback (0.011 vs. 0.013 N respectively), but the .5 N target (0.29 N) having less variability than the 1 N target (0.49 N) in the no vision condition. Post hoc analysis revealed that the 0.5 N (0.02 N) target had less variability than the 1 N target (0.031 N) which had less variability than the 2 N target (0.057 N). As expected the no vision condition had greater variability than the vision condition (0.055 N vs. 0.17 N, respectively).

Statistical analysis of force RMSe revealed a main effect of force level \([F(2,20) = 107; \ p < .05, \ \eta^2 = 0.92]\), visual feedback \([F(1,10) = 90.6; \ p < .05, \ \eta^2 = 0.91]\), and an interaction between force level and visual feedback \([F(2,20) = 52.7; \ p < .05, \ \eta^2 = 0.85]\). The interaction between force level and vision resulted from no effect of the vision removal at the 0.5 N (0.04 vs. 0.06), but a significant increase in RMSe with the removal of vision in the 1 (0.11 vs. 0.31 N) and 2 N (0.19 vs. 0.36) tasks. Post hoc analysis revealed that RMSe increased with each target force level (0.04, 0.21 and 0.30 N, respectively). The no vision condition had significantly greater RMSe compared with the vision condition (0.15 vs. 0.22).

Figure 2A depicts a representative subject’s force output to a 1 N target in the no vision condition and the corresponding beat to beat blood pressure oscillation. There appears to be a relation between blood pressure oscillation and fluctuations in force output. Figure 2B is coherence spectrum between force output and blood pressure oscillation from the same data. Multiple peaks are apparent within the spectra with dominant peaks at approximately 4, 11 and 19 Hz. To quantify how force level and visual condition impacted the association between force output and peripheral blood pressure oscillation, maximum coherence and the frequency at which it occurred was calculated. Average maximum coherence (0.43) was significantly different than zero \((p < .05) and ranged from 0.26 to 0.84. Statistical analysis revealed a main effect for vision \([F(1, 10) = 4.9; \ p < .05, \ \eta^2 = 0.33]\), which resulted from greater coherence in the vision condition compared with the no vision condition (0.45 vs. 0.40, respectively) (See Figure 3A).

Frequency of maximum coherence on average occurred within the 8–12 Hz bandwidth (See Figure 3B). Statistical analysis revealed an interaction between force level and visual condition \([F(2,18) = 3.9; \ p < .05; \ \eta^2 = 0.30]\). It was found that this interaction resulted from the .5 N target having lower frequency of maximum coherence in the vision condition (7.48 Hz) than the 1 N target (11.37 Hz), but greater frequency in the no vision condition (12.75 Hz vs. 8.38 Hz, respectively).

It is important to note that although the frequency of maximum coherence occurred on average within the 8–12 Hz bandwidth, there was considerable amount of variance in the bandwidth peak coherence occurred. A visual examination of the distribution of the bandwidth of peak coherence revealed that there were three
Figure 2 — A) Representative force output in the 2 N no vision condition with the corresponding beat to beat blood pressure oscillation. B) The corresponding coherence spectrum between force output and blood pressure oscillation. The solid line indicates the coherence spectrum, while the dashed line illustrates the 95% confidence interval.
Figure 3 — A) Maximum coherence (+ standard error) of blood pressure oscillations and force output as a function visual feedback. B) Frequency of maximum coherence (+standard error) between blood pressure oscillations and force output as a function of force level and visual feedback. * p < .05
main clusters: one centered within the 0–6 Hz bandwidth, an 8–14 Hz cluster and a 17–23 Hz cluster. 43.3% of trials were found to have a peak coherence occur in the low frequency bandwidth (0–6 Hz), 38.3% in between 8–14 Hz bandwidth and the remaining 18.3% in the 17–23 Hz bandwidth. There was no significant relation between frequency of peak coherence and maximal coherence ($p > .05$).

**Discussion**

The purpose of this investigation was to examine whether there is a contribution of CBI to fluctuations in continuous isometric force output in healthy young adults. Overall, it was found that there was a significant coherence between force fluctuations and blood pressure oscillations. This coupling was found to occur on average in the 8–12 Hz frequency range. The coherence between force fluctuations and blood pressure was found to be mediated by the availability of visual feedback, but not force level.

The 8–12 Hz bandwidth is the frequency range which is traditionally attributed to physiological tremor. Physiological tremor results from a complex interaction of neural and mechanical processes (Marsden, 1984). Marsden and colleagues (1969) demonstrated that cardioballistics accounts for ~10% of physiological tremor amplitude. Although this finding has been replicated (Elble & Randall, 1978), there is some confusion regarding its universality. It has been argued that the CBI contribution to tremor genesis varies across individuals and only significantly contributes to tremor amplitude in 30% of subjects (Wade, Gresty, & Findley, 1982). In contrast to this proposition, all the subjects in the current investigation demonstrated significant coherence between force fluctuations and blood pressure oscillations across all trials. Overall, the observation that coherence between beat to beat blood pressure and force was greatest in the 8–12 Hz bandwidth is congruent with the proposition that CBI contributes to fluctuations in force output.

Interestingly there was no change in the strength of the coupling between force output and blood pressure as a function of force level. Since a limited range of force levels were examined (~5–20% MVC), it is not clear if the contribution of the cardioballistic impulse would be altered at higher force levels. Gallasch and Kenner (1997) suggested that there was a decrease in the cardioballistic contribution to force fluctuations during increasing exertion in a grip task—but only visible assessment of waveforms was used.

It is maintained that force variability scales to force level because of the differential influence of various neuromuscular mechanisms across the working range of force output (Christou et al., 2002; Sosnoff, Valentine & Newell, 2006; Taylor et al., 2003). The lack of effect of force level suggests that the contribution of CBI to fluctuations in force output is invariant and distinct from neuromuscular mechanisms that drive the scaling of force variability to magnitude of force level. It is proposed that CBI contributes to the baseline variability to force fluctuation and as such is a limiting factor in precise force output—that is the amount of variability which is independent of muscle activation dynamics. Further work is needed to test this proposition.

It is well known that force output with visual feedback is less variable than without feedback (Slifkin et al., 2000; Sosnoff & Newell, 2005; cf Tracy et al., 2007). It was predicted that the coupling between blood pressure and force output
would be greater in the no vision condition. However, the exact opposite was observed: visual feedback resulted in greater coherence between force and blood pressure oscillations. Since blood pressure oscillations occurred at the same frequency across the experiment, the perturbation due to cardioballistics is rhythmic and easily predictable. It is proposed that trials with visual feedback give subjects a greater chance to observe CBI and entrain the force output with the cardioballistic impulse. Consequently, there is a greater increase in coherence between the signals. When considering this potential visual effect on CBI on force variability, one should keep in mind that the functional significance of a .05 difference in coherence between vision and no-vision conditions is not clear. Readers should keep in mind that statistical significance is not equivalent to functional/clinical significance.

The focus of the current investigation was to assess the contribution of CBI in the form of local blood pressure oscillations to force variability in healthy young adults. It is not clear if the findings generalized to other populations. Several conditions such as advanced aging, hypertension, diabetes, etc. functionally and structurally change the cardiovascular system (Lakatta, 2000). For instance, hypertension, commonly referred to as high blood pressure, is common adverse age-related disease of the cardiovascular system. Elevated blood pressure results, in part, from increased cardiac contractility to eject blood into a less compliant vascular system. It is logical to speculate that increases in the force causing blood to be ejected into the arterial system would result in a greater CBI contribution to force variability. Further research is warranted to examine this possibility.

Although this investigation provides unique information, a limitation is that it does not experimentally manipulate BP oscillations. Manipulations of BP oscillations would confirm the effect of CBI on force fluctuation. However, manipulations of BP have inherent health risks that make it less appealing in a nonhealth care setting. In addition, only a small portion of potential force levels were used. As such the association between blood pressure and force fluctuations is not clear.

The novel contribution of this investigation is that there is a significant coupling between local blood pressure oscillations and force fluctuations in the 8–12 Hz bandwidth in healthy young adults. Since this coupling was not influenced by increasing force magnitude, it is proposed that the peripheral alterations in blood pressure are a small, albeit significant contributing factor in fluctuations in constant isometric force output.

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References


