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Use of Whole Body Vibration in Individuals with Chronic Stroke: Transmissibility and Signal Purity.

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Abstract: This study examined (1) the influence of whole body vibration (WBV) frequency (20Hz, 30Hz, 40Hz), amplitude (low: 0.8mm and high: 1.5 mm) and body postures (high-squat, deep-squat, tip-toe standing) on WBV transmissibility and signal purity, and (2) the relationship between stroke motor impairment and WBV transmissibility/signal purity. Thirty-four participants with chronic stroke were tested under 18 different conditions with unique combinations of WBV frequency, amplitude, and body posture. Lower limb motor function and muscle spasticity were assessed using the Fugl-Meyer Assessment and Modified Ashworth Scale respectively. Nine tri-axial accelerometers were used to measure acceleration at the WBV platform, and the head, third lumbar vertebra, and bilateral ankles, knees, and hips. The results indicated that WBV amplitude, frequency, body postures and their interactions significantly influenced the vibration transmissibility and signal purity among people with chronic stroke. In all anatomical landmarks except the ankle, the transmissibility decreased with increased frequency, increased amplitude or increased knee flexion angle. The transmissibility was similar between the paretic and non-paretic side, except at the ankle during tip-toe standing. Less severe lower limb motor impairment was associated with greater transmissibility at the paretic ankle, knee and hip in certain WBV conditions. Leg muscle spasticity was not significantly related to WBV transmissibility. In clinical practice, WBV amplitude, frequency, body postures need to be considered regarding the therapeutic purpose. Good contact between the feet and vibration platform and symmetrical body-weight distribution pattern should be ensured.

Keywords: whole body vibration; stroke; rehabilitation; transmissibility.

1. Introduction

Whole-body vibration training (WBV) has become increasingly popular as an exercise modality (Rittweger, 2010). It is believed that vibration stimulus can activate muscle spindles causing alpha-motoneurons excitation, thus augmenting muscle contraction (Luo et al., 2005; Ribot-Ciscar et al., 1998) and motor unit synchronization (Jackson and Turner, 2003). As bone tissue is very sensitive to mechanical loading, the dynamic mechanical loading involved in WBV training would also enhance bone anabolic response (Torcasio et al., 2008). However, significant reduction of touch-pressure sensation on the sole of the foot and balance ability was found immediately after exposure to vibration (Sonza et al., 2015). Long-term occupational exposure to high-magnitude vibrations may lead to deleterious effect in the spinal and reproductive systems (Bovenzi, 2006). Excessive transmission of vibration to the head could also cause damage to the retina (Ishitake et al., 1998) and inner ear (Bochnia et al., 2005).

During WBV treatment, the vibration is transmitted from the vibration source (e.g., vibration platform) to the target body part(s) (Luo et al., 2005). However, the biodynamic responses to vibration depend on many factors, including the vibration amplitude and frequency of platform (Matsumoto and Griffin, 1998), body posture (Lafortune et al., 1996), muscular activation(Tarabini et al., 2014) and musculoskeletal compliance (Wakeling et al., 2002). Thus, the vibration characteristics (e.g., magnitude) would vary in different body parts (Matsumoto and Griffin, 1998) and the response of the human body to the platform vibration is non-linear (Lafortune et al., 1996; Matsumoto and Griffin, 1998). In addition, there may be a distortion of sinusoidal waveforms as the signals are transmitted upward. The degree of signal attenuation (i.e., transmissibility) and how well the original signal waveforms are retained during transmission (i.e., signal purity) may have major impact on therapeutic efficacy (Kiiski et al., 2008). Moreover, transmissibility to the head and resonance effect should be minimized for safety concerns (Caryn et al., 2014; Kiiski et al., 2008). To this end, the ratio of acceleration measured at the level of the vibration platform and that at the target body part, have been used as a measure of transmissibility (Matsumoto and Griffin, 1998). The site-specific vibration frequency profile (i.e., signal purity) can be evaluated using vibration spectral analysis (Griffin, 1996; Kiiski et al., 2008).

Previous research on vibration transmissibility was only conducted in healthy young adults (Abercromby et al., 2007; Avelar et al., 2013; Caryn et al., 2014; Cook et al., 2011; Crewther et al., 2004; Kiiski et al., 2008; Lafortune et al., 1996; Matsumoto and Griffin, 1998; Rubin et al., 2003; Tankisheva et al., 2013). Generally, the signals are attenuated as they are transmitted from the feet upward to other body sites (Cook et al., 2011; Harazin and Grzesik, 1998; Tankisheva et al., 2013). Increased frequency was found to be associated with lower vibration transmissibility (Caryn et al., 2014; Cook et al., 2011; Kiiski et al., 2008). However, amplification of signals can be found in certain anatomical landmarks (Kiiski et al., 2008; Matsumoto and Griffin, 1998), particularly when the frequency was less than 20Hz (Crewther et al., 2004; Kiiski et al., 2008; Matsumoto and Griffin, 1998; Tankisheva et al., 2013). Changes in posture could also modify the vibration transmission. However, only the effect of knee flexion angle was studied in previous research (Abercromby et al., 2007; Avelar et al., 2013; Cook et al., 2011; Rubin et al., 2003; Tankisheva et al., 2013). The interaction effects of vibration frequency, amplitude, and postures on transmissibility was reported in one study, but transmissibility was measured at the head only (Caryn et al., 2014). Moreover, WBV signal distortion during transmission is very understudied. Only Kiiski et al. (2008) have examined WBV signal purity during erect standing in healthy young adults. Signal purity in other postures that are often used in WBV exercise training has not been studied.

Although WBV has been widely used and examined among people with stroke (Brogardh et al., 2012; Chan et al., 2012; Liao et al., 2016; Marin et al., 2013; Miyara et al., 2014; Pang et al., 2013; Tankisheva et al., 2014; Tihanyi et al., 2007; van Nes et al., 2006), mixed results on various clinical outcomes are found. The discrepancies in results may be due to the difference in WBV protocols used. Thus, it is important to conduct more fundamental research on transmissibility and signal purity in the stroke population. The information gained would inform the design of WBV protocols, which can be formally tested in efficacy studies. Stroke survivors may have altered musculoskeletal characteristics (Cruz et al., 2009). Moreover, the muscle activation patterns with WBV stimulation among individuals with stroke were also different from healthy young adults (Liao et al., 2014b). Because the musculoskeletal system is a major pathway through which the WBV is transmitted, it is highly likely that WBV transmissibility in stroke patients could be very different from that in able-bodied individuals. However, to date, the research on vibration transmission and signal purity for the stroke population is lacking.

The objectives of this study were (1) to investigate the influence of WBV frequency, amplitude, and body posture on WBV transmissibility and signal purity; and (2) to examine the impact of stroke motor impairment and spasticity on WBV transmissibility. It was hypothesized that (1) WBV transmissibility and signal purity would be influenced by vibration frequency, amplitude, body postures, and their interactions; (2) WBV transmissibility and signal purity would demonstrate a significant difference between the paretic and non-paretic side; and (3) WBV transmissibility would be associated with the motor impairment level or muscle spasticity of the affected lower extremity.

2.Method

2.1 Subjects

Individuals with chronic stroke were recruited from the local community using convenience sampling. The inclusion and exclusion criteria are shown in supplementary table 1. The study was approved by the Human Subjects Ethics Sub-committee of the University and conducted according to the Declaration of Helsinki. Written informed consent was obtained from each participant before data collection.

2.2 Experimental setting

This was an experimental study with repeated measures. A WBV device (Fitvibe Excel, GymnaUniphy, Belgium) generating sinusoidal vertical vibrations was used. Eighteen WBV conditions generated by different combinations of two vibration amplitudes (low: 0.8mm; high: 1.5mm), three vibration frequencies (20Hz, 30Hz, 40Hz), and three postures (high-squat with knee flexion at 30°, deep-squat with knee flexion at 60°, tip-toe standing) were tested. The specific requirements for each posture are described in Figure 1. The rationale of the above protocols is explained in supplementary table 2. Due to safety, all participants were asked to hold the handrail lightly for balance only. Each trial was about 20 seconds. The 18 trials were conducted in randomized order with a 1-minute rest period between each trial. Two researchers provided standby supervision to ensure safety and correct postures (e.g., correct knee angle and no trunk/head movement). The trial would be terminated immediately if any adverse symptoms (e.g., fatigue, dizziness) were reported. All the experimental procedures were completed in the same laboratory of the university. The session was about 1.5 hours, including the set-up time and rest periods.

2.3 Measurements

The demographic information was obtained first by interviewing the participants. Motor function of the paretic leg was assessed by a physiotherapist using Fugl-Meyer Assessment (FMA). Higher scores denoted less impaired motor function (Sullivan et al., 2011). The spasticity of ankle plantar flexors, knee flexors, and extensors was also rated using the Modified Ashworth Scale (MAS), and a higher score indicated more severe spasticity (Bohannon and Smith, 1987).

Seven tri-axial accelerometers (Dytran Model 7523A5, Chatsworth, Canada) were mounted on tailor-made polyester board (weight 10g; Otto Bock, Duderstadt, Germany) using screws, which were attached to the skin overlying the specific body sites using tapes (Omnifix® elastic, Hartmann-Conco, Heideman, Germany) and self-adherent wraps (Coban®, 3M, Saint Paul, US): bilateral medial malleolus (ankle), bilateral medial condyle of the femur (knee), bilateral greater trochanter (hip), and third lumbar vertebra (L3) (Matsumoto and Griffin, 1998). To measure the acceleration at the head, one tri-axial accelerometer was fixed on the disposable dental impression that was held firmly between the upper and lower teeth (Harazin and Grzesik, 1998). Another tri-axial accelerometer was attached to the center of WBV platform to measure platform acceleration. Before data collection, the orientation of accelerometers was calibrated and checked to secure properly alignment of axis coordinates.

2.4 Data processing

Acceleration signals were recorded and digitized via a 32-channel analog-to-digital converter (Model DT9844, Data Translation, Norton, US) using a custom program written in LabView (version 8.6, National Instruments, Austin, US) with a resolution of 20-bit and sampling frequency of 1000Hz.

Offline data analysis was performed using a custom-written script in MATLAB (version R2014b, MathWorks, Natick, US). For each 20-second trial, the data from one 3-second period in the middle section were selected for analysis (Kiiski et al., 2008). This single 3-second period should contain 40-120 vibration cycles, depending on the vibration frequency. The DC offset induced by the gravitational acceleration was removed. In the time domain, resultant accelerations were calculated from the vector sum of x,y, z-axis, which represented the absolute acceleration of each vibration trial. The root-mean-square (RMS) value of the resultant acceleration was calculated in a 3-s window for each vibration trial (Caryn et al., 2014). The transmissibility of vibration to each body site was calculated as the ratio of the acceleration RMS of the site-specific signal to the acceleration RMS of the WBV platform (Kiiski et al., 2008). A transmissibility ratio larger than 1.0 indicated amplification of vibration signals transmitted from the platform to the measured body site, while the ratio less than 1.0 indicated dampening of signals during its transmission.

For each trial, the frequency domain of the acceleration signals at the platform, and each body site was also analyzed by performing fast Fourier transform. The proportion of signal power with ±1Hz of the vibration frequency at the level of the platform (i.e., nominal frequency) was computed to assess the signal purity of the sinusoidal waveform using the following formula (Kiiski et al., 2008):

$$\text{Signal Purity}=\frac{Px(nominal\ frequency\pm1)+Py(nominal\ frequency\pm1)+Pz(nominal\ frequency\pm1)}{Px+Py+Pz}\times 100\%$$

in which the Px (nominal frequency±1Hz), Py (nominal frequency±1Hz), Pz (nominal frequency±1Hz) indicated the power of nominal frequency±1Hz in x, y, and z-axis respectively; and Px, Py, and Pz indicated signal power at the level of the platform in x, y, and z-axis respectively (figure 2). The greater the signal purity value, the better the sinusoidal waveforms were maintained. A value of greater than 80% was considered to be adequate (Kiiski et al., 2008).

2.5 Statistical analysis

Statistical analysis was conducted using SPSS (version 22, IBM, Armonk, NY). The significant level was set at $p<0.05$. Normality was checked by Shapiro-Wilk test. Two separate three-way repeated-measures ANOVA (within-subject factors: three frequencies, two amplitudes, three postures) was performed to investigate the effect of vibration frequency, amplitude and body postures and their interactions on signal transmissibility and purity, respectively. The Greenhouse-Geisser epsilon adjustment was used when sphericity assumption was violated. Post-hoc analysis using paired-t Bonferroni adjustment was performed if any overall significant results were identified. The effect size was expressed as partial eta squared ($ŋ_p^2$) (Fritz et al., 2012). Next, paired t-tests with Bonferroni adjustment were used to compare the difference in vibration transmissibility and signal purity between the paretic and non-paretic side for each WBV condition. Spearman’s rho was used to examine the relationship between (a) FMA score, (b) ankle plantarflexors MAS score, (c) knee flexor MAS score, (d) knee extensor MAS score and transmissibility on the paretic side.

3. Results

Thirty-four participants with chronic stroke [12 women, 22 men; mean age (SD): 62.3 (2.7) years] completed all testing. No adverse effect (e.g., dizziness, nausea) was reported during our study. The demographics are summarized in Table 1. The acceleration-RMS generated by the platform ranged from 0.79*g* to 4.94*g* ($1g \approx 9.81 m/s^2$). The signal purity measured at the level of the platform ranged from

93.6% to 96.8%, indicating that the WBV platform was adequate in generating the intended sinusoidal signals as the signal purity value was greater than 80% (Kiiski et al., 2008).

3.1 Transmissibility

The transmissibility in each measured body site under various vibration settings is illustrated in Table 2. Amplification of vibrations was observed at the ankle on both sides (transmissibility: 1.05-1.98).

The transmissibility decreased with increasing frequency in all sites except the ankles ($p<0.001$) (Figure 3). The significant frequency × posture interaction effect was found in all measured body sites ($F=4.01\text{-}56.54$, $p\leq0.009$, $\eta_p^2=0.11\text{-}0.63$). The frequency × posture interaction effect was more prominent with high-amplitude WBV ($\eta_p^2=0.744$) comparing to the low-amplitude WBV ($\eta_p^2=0.538$) at the knee bilaterally, thus accounting for the significant frequency × posture × amplitude interaction effects at these two sites ($F=3.35\text{-}3.97$, $p=0.012\text{-}0.031$, $\eta_p^2=0.09\text{-}0.11$). The transmissibility decreased with increasing amplitude in all sites except the ankles ($p\leq0.02$)(Figure 4).

3.2 Signal purity

The mean signal purity values are shown in Table 3. In general, the signal purity was satisfactory (≥80%), with a few exceptions. Frequency spectrum analysis revealed high-frequency components at the ankles when 20Hz and high amplitude was used (Figure 5), which accounted for a relatively low signal purity value (73%). Signal purity was well below 80% at the paretic hip when 40Hz was used during high squat and tip-toe standing, and at the head during high squat. WBV signal purity showed no significant frequency × posture × amplitude interaction effects in all measured sites. Frequency × posture interactions were significant at bilateral ankles, knees and hips

(F=2.84-4.00, $p \leq 0.011$, η_p^2=0.12-0.18) (Figure 6). Amplitude × posture interactions were significant in all measured sites (F=9.40-40.00, $p \leq 0.001$, η_p^2=0.22-0.55) (Figure 7).

3.3 Comparison between the paretic and non-paretic side

On the non-paretic side, there was significantly greater transmissibility in the ankle during most tip-toe standing conditions when compared with the corresponding values on the paretic side. There was no significant difference between the paretic side and non-paretic side regarding signal purity, regardless of the WBV conditions.

3.4 Relationship to motor impairment and spasticity

There was no significant association of the MAS score of the knee flexors/extensors and ankle plantarflexors with transmissibility. The FMA score was positively associated with transmissibility values measured at the paretic ankle during most tip-toe standing conditions (rho=0.36-0.60, $p \leq 0.039$). The FMA score was also positively associated with the transmissibility (rho=0.34-0.38, $p \leq 0.028$) and signal purity (rho=0.38-0.47, $p \leq 0.010$) at the paretic knee and hip when performing a deep squat and tip-toe standing (40Hz/high amplitude).

4. Discussion

Our hypothesis was confirmed that WBV transmissibility and signal purity were influenced by vibration frequency, amplitude, body postures and their interactions. Generally, above the ankle, increased vibration frequency or amplitude led to deceased WBV transmissibility, although the magnitude of the trend was influenced by body posture. Postures and body segments were the main factor that influenced vibration transmissibility (Harazin and Grzesik, 1998). Bending motion of the knees with the rotational motion of the pelvis could contribute the attenuation of vibration in the upper body (Matsumoto and Griffin, 1998). Therefore, lower transmissibility above the knee joint was observed in the deep squat posture compared to the high-squatting posture. This finding is largely in line with previous literature on healthy adults (Abercromby et al., 2007; Avelar et al., 2013; Rubin et al., 2003; Tankisheva et al., 2013).

In tip-toe standing, the relationship between vibration transmissibility and frequency showed a distinct pattern. Except at the head, the vibration transmissibility was significantly lower when assuming tip-toe standing compared with the two squat positions, which confirmed the speculation that more weight bearing on the forefoot when standing could lead to lower transmissibility (Rittweger, 2010; Tankisheva et al., 2013). Muscle activation induced by the vibration could attenuate the vibration energy (Tankisheva et al., 2013; Wakeling et al., 2002) and may alter the vibration transmissibility in body parts above the muscles activated. During tip-toe standing, the base of support was relatively small, which challenged postural stability. Muscle activity of the lower limbs may be increased to maintain the posture (Branthwaite et al., 2012), which in turn led to more substantial attenuation of the vibration energy (Branthwaite et al., 2012; Liao et al., 2014a). In addition, the foot arch created during tip-toe standing could further absorb the signals (Simkin et al., 1989). However, the transmissibility to the head was greater during tip-toe standing compared with deep squat, probably because the increased trunk rigidity related to augmented activation of trunk muscles (Branthwaite et al., 2012) and pressure to torso joints during tip-toe standing (Marouane et al., 2015).

Vibration amplification (i.e., transmissibility >1.0) was observed at the ankles. The power spectrum analysis also showed that higher frequency component was generated (figure 5). A similar phenomenon was previously found in healthy young adults (Harazin and Grzesik, 1998). It was believed that during vibration cycles, the feet might lose contact with vibration platform that led to an air-borne phase for the feet (Rittweger, 2010). As a result, the collision occurred between platform and feet, which may generate impact forces (Rittweger, 2010) and thus led to excessive loading in the ankle (Kiiski et al., 2008).

In contrast to our hypothesis, the non-paretic side showed greater transmissibility than the paretic side only at the ankle during tip-toe standing. Tip-toe standing, with a narrow base of support, was quite challenging for stroke patients. The participants may put more weight on the non-paretic side to improve stability, which resulted in lower transmissibility on the paretic side. This may also partly explain why better motor recovery of the paretic leg (i.e., higher FMA score) was correlated with greater transmissibility on the same side, as those with better motor recovery may have better ability to weight bear on the paretic side. Spasticity, on the other hand, was measured while the participants were in a resting state, and thus may not have a noticeable influence on the transmissibility of vibrations.

When setting WBV training protocol, safety should be the key consideration. Excessive vibration loading to the head should be avoided (Caryn et al., 2014) especially for people with stroke. In our study, although the average vibration intensity measured at the platform was up to 4.94g, the vibration signals were substantially attenuated at the level of the head (transmissibility ratio: 0.04-0.30; vibration intensity: 0.11-0.60g).

To achieve the desired therapeutic effect on bone and muscle, adequate vibration transmissibility and signal purity at the target treatment site is necessary (Pang et al., 2013). On the other hand, low vibration transmission to the head is important to ensure safety. Based on our study findings (Table 2; Table 3), a combination of WBV frequency at 20Hz, low amplitude and deep squatting position may be a better choice if the treatment goal was to enhance/maintain bone mass in the hip and lumbar spine. This concurred with the finding of a meta-analysis which showed low-frequency (20Hz) and high-magnitude (>1g) vibrations could lead to significant increase of hip and lumbar bone density in healthy older adults (Oliveira et al., 2016). As our accelerometers were put in bony body area rather than the muscles, vibration loading in muscle could not be directly detected. However, as muscle activation would have a major damping effect on vibration signals (Wakeling et al., 2002), lower transmissibility may indicate more muscle activation below the measured bony sites. Thus, to activate leg muscles, deep squatting with WBV at 30-40Hz would be more appropriate. However, the above hypotheses will need to be tested in future research, with measurement of muscle activity/strength and bone quality.

As the aim was to examine the transmissibility and signal purity, the duration of exposure to WBV per testing condition was very brief (20 seconds). The period should generate an adequate number of vibration cycles for our data analysis. To study the therapeutic or harmful effect of such exposure, probably a longer period of exposure is required and will need further investigation. Although some of the testing conditions involved high-intensity WBV and hence may induce a high level of muscle work, there should not be major concerns with muscle fatigue or damage. First, we did monitor closely the patients' condition throughout the experiment. No limb numbness, discomfort or muscle fatigue was reported by our subjects. Second, the actual level of muscle activation may not be very high. Previous studies examining the effect of WBV on muscle activation in stroke patients (Liao et al., 2014a; Liao et al., 2015) showed that addition of vibration (20-30Hz, 0.44-0.60mm, 0.96g-1.61g) led to an increase in muscle activity by 10-25% of the maximal voluntary contraction. The levels of muscle activation attained did not exceed 40% maximal voluntary contraction while assuming various postures that were similar to those used in our study, even with their high-intensity protocol (1.61g) (Liao et al., 2014a; Liao et al., 2015). There was no further increase in muscle activation levels for the majority of muscle groups as the intensity was changed from 0.96g to 1.61g, indicating a possible saturation in muscle response (Liao et al., 2014a; Liao et al., 2015). There was also no significant difference in WBV-induced EMG response between the affected and unaffected side (Liao et al., 2014a; Liao et al., 2015). Moreover, we found that more severe motor deficit was associated with lower transmissibility. Taken together, it is unlikely that the protocol used here would induce a very high level of muscle activation and cause muscle damage. In fact, resistance training in stroke patients often involves a high level of muscle work (60-80% of 1 repetition maximum) in order to induce strength training effect (Patten et al., 2004). Tankisheva et al. (2014) also found that high-intensity vibrations (frequency: 35Hz-40Hz, amplitude: 1.7-2.5mm, intensity: up to 16.1g) could significantly increase knee extensor strength after 6 weeks of training without causing any fatigue or pain.

Vibration amplification was found in the ankle joint at 20-40Hz and knee joint at 20Hz. Therefore, caution should be exercised when applying WBV to stroke patients who also have ankle or/and knee pathology, especially when the above frequencies are used. For people with severe stroke motor impairments, uneven weight bearing would occur during WBV. Hence, good contact between the feet and vibration platform and more symmetrical body-weight distribution pattern should be ensured (Emerenziani et al., 2014), thus to reduce the air-borne effect at the ankles and promote better vibration transmission to the paretic side (Rittweger, 2010).

Several limitations of our study should be considered. First, skin-mounted and bite-bar accelerometers were used. Although it may not be as accurate as the bone-mounted method, the non-invasive skin-mounted or bite-bar accelerometers provide a much safer and feasible option, and have been used in previous studies (Abercromby et al., 2007; Avelar et al., 2013; Caryn et al., 2014; Cook et al., 2011; Crewther et al., 2004; Harazin and Grzesik, 1998; Kiiski et al., 2008; Lafortune et al., 1996; Matsumoto and Griffin, 1998; Tankisheva et al., 2013). We also put light-weight plastic between the skin and accelerometer to minimize the influence of skin stretch, uneven bony surface, temperature and humidity (Matsumoto and Griffin, 1998). Our power spectrum analysis also showed that the signal purity could reach 96.5%, 94.4%, 87.0%, 92.,2% and 87.3% in ankles, knees, hips, lumbar and head, respectively, which suggested that our accelerometers were well mounted. Second, we did not report how muscle activation varied with the different WBV parameters tested in the current study. Future studies may explore the muscle activation and its relationship with the WBV transmission. We also did not assess the beneficial or harmful effects of long-term exposure of WBV. This issue awaits further research.

In summary, WBV amplitude, frequency, body postures and their interaction could significantly influence the vibration transmissibility and signal purity among people with chronic stroke. Leg muscle spasticity was not significantly related to WBV transmissibility. Less severe lower limb motor impairment was associated with greater transmissibility at the paretic ankle, knee, and hip in certain WBV conditions. In clinical practice, WBV amplitude, frequency, body postures need to be considered regarding the therapeutic purpose. Good contact between the feet and vibration platform and symmetrical body-weight distribution pattern should be ensured.

Complete with Interest

The authors declare that they have no conflict of interest.

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