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The difference between first and second harmonic amplitudes correlates between glottal airflow and neck-surface accelerometer signals during phonation

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Abstract: Miniature high-bandwidth accelerometers on the anterior neck surface are used in laboratory and ambulatory settings to obtain vocal function measures. This study compared the widely applied L_1–L_2 measure (historically, $H1$–$H2$)—the difference between the log-magnitude of the first and second harmonics—computed from the glottal airflow waveform with L_1–L_2 derived from the raw neck-surface acceleration signal in 79 vocally healthy female speakers. Results showed a significant correlation ($r = 0.72$) between L_1–L_2 values estimated from both airflow and accelerometer signals, suggesting that raw accelerometer-based estimates of L_1–L_2 may be interpreted as reflecting glottal physiological parameters and voice quality attributes during phonation.

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1. Introduction

Neck-surface accelerometers have been used for several decades to yield measures of laryngeal voice production that can supplement or supplant acoustic voice measures.[1] In such applications, accelerometers offer advantages over microphones by being much less sensitive to environmental noise sources (including other speakers) and, when placed below the larynx, are also not sensitive to the articulatory modulation of the speech acoustic signal (thus, alleviating confidentiality concerns).[2] In addition, the low profiles of accelerometer sensors are conducive for use in ambulatory monitoring devices that are worn throughout an individual's daily activities to capture/characterize typical and atypical vocal behavior.[3–5] In addition to sensing phonatory characteristics, accelerometers have also been employed to monitor subglottal resonances,[6] nasal resonances,[7] and chest wall vibrations during singing.[8]

When positioned on the anterior neck surface below the larynx, accelerometers capture phonation-related vibration (skin surface acceleration) that is related to both tissue-to-tissue transmission of vocal fold collision forces and air-to-tissue transmission of aerodynamic energy through the trachea.[9] Early applications took advantage of the noise-robustness of accelerometers to track fundamental frequency in high-noise environments.[10] More recent studies have expanded the suite of measures extracted from the accelerometer signal to include temporal, spectral, and cepstral measures of voice production. For example, the field of voice dosimetry[11,12] has taken advantage of the long-term, wearable nature of accelerometers to estimate vocal dose measures that combine estimates of voicing duration, sound pressure level, and fundamental frequency to quantify the risk of developing certain types of voice disorders.[13] The extraction of spectral and cepstral measures from the accelerometer signal has been motivated by the desire to quantify changes in vocal function over the course of an individual's day and to study phonatory characteristics of patients with voice disorders and vocally healthy speakers.[3,14]

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Relationships between traditional acoustic-based estimates of vocal function and their skin acceleration-based counterparts have been investigated to gain additional insight into the interpretability of accelerometer-based vocal parameters. High values for Pearson's correlation coefficient have been found when comparing temporal measures ($r > 0.99$ for average fundamental frequency, $r = 0.80$–0.91 for cycle-to-cycle jitter) extracted from the acoustic and acceleration signals of sustained vowels.[15] Similarly, strong correlations have been found between microphone- and accelerometer-based estimates of cepstral peak prominence ($r = 0.82$–0.90), a measure of overall periodicity and harmonics-to-noise ratio. In contrast, weaker correlations were exhibited by amplitude-based measures of cycle-to-cycle shimmer ($r = 0.15$–0.39), harmonics-to-noise ratio ($r = 0.14$–0.70), and spectral tilt across eight harmonics ($r = 0.05$–0.59). The low correlation between microphone- and accelerometer-based estimates of spectral tilt were potentially due to the high-bandwidth nature of the measure that was especially sensitive to the spectral compensation of three formants[16] interacting with the spectral resonances/nulls of the pneumotachograph mask.

The current study examines a low-bandwidth measure of spectral tilt that can be directly estimated from the accelerometer signal—the spectral amplitude of the first harmonic relative to the amplitude of the second harmonic. Computed in units of decibels (dB), the difference between these first two harmonic levels is often termed $H1$–$H2$ in the literature[17] but, following a recent consensus,[18] the measure is notated here as L_1–L_2 to clarify that harmonic levels (and not frequencies or other properties) are employed. When computed from an acoustic microphone signal, L_1–L_2 is a common measure of spectral tilt that has been related to glottal airflow pulse skewness, open quotient, and medial vocal fold thickness.[19–22] Pulse skewness and spectral tilt of the glottal airflow waveform have been implicated as correlates of breathy[23] and strained voice quality.[24] For example, a glottal vibratory pattern that exhibits reduced vocal fold contact and produces breathier voice quality is associated with higher first-harmonic energy in the acoustic domain.[21] Thus, along with other measures associated with voice quality, estimates of L_1–L_2 computed from the voice source can provide valuable information about underlying phonatory physiologic mechanisms[25] and also has good potential for use in the clinical assessment and treatment (e.g., biofeedback during voice therapy) of voice disorders.

L_1–L_2 values from neck-surface accelerometer signals are hypothesized to correlate with L_1–L_2 values measured traditionally in the acoustic and airflow domains because of the underlying theory that neck-surface vibration reflects the radiation of the voice source signal through the trachea (including the influence of the tracheal resonances) and anterior neck tissue.[26] Acoustically, measures of the first harmonic amplitude have been normalized after subtracting either the second harmonic amplitude, first formant amplitude, third formant amplitude, or overall sound pressure level of a voiced segment.[17,21] Because formant amplitude estimates depend on the harmonic sampling of the underlying spectral envelope, formant amplitudes and overall sound pressure level (also driven by the formant amplitudes) are less desirable as references for the first harmonic amplitude.[21] Thus, the second harmonic amplitude is the reference chosen in the current study, and L_1–L_2 represents a low-bandwidth estimate of the harmonic decay of the voice source. Additional correction factors are often applied to the harmonic amplitudes to remove the effects of the spectral envelope imposed by formants to characterize the voice source.[16,17] Although similar correction factors for the harmonic amplitudes of the accelerometer signal could be performed, the need to compensate for the amplification/attenuation effects of subglottal resonances is less pronounced due to the relative time invariance of subglottal resonances relative to supraglottal formants across different vowel types.

The purpose of this study was to investigate the relationship between L_1–L_2 values computed from the raw (unfiltered) neck-surface accelerometer signal and L_1–L_2 values from the reference glottal airflow waveform obtained after standard inverse filtering of the oral airflow. A high correlation between L_1–L_2 values in the two domains would provide evidence for the interpretation of accelerometer-based L_1–L_2 as a correlate of glottal closure properties (e.g., open quotient) and voice source pulse skewness.

2. Methods

The study sample consisted of 79 adult female speakers with no history of voice disorders who were recruited to participate in a larger study of ambulatory voice monitoring.[3] The normal vocal status of the speakers was verified via interview and a laryngeal stroboscopic examination. Only data from female participants were analyzed since they were recruited as matched controls for patients with voice disorders that exhibit a higher prevalence in female speakers (thus, more data acquired from female subjects). The mean [standard deviation (SD)] age of the participants was 29.6 (13.0) years.

The data acquisition protocol was based on past aerodynamic studies[27] and detailed in recent work.[28] Participants were asked to produce three sets of five consecutive /pae/ syllables for each of three different loudness conditions (soft, comfortable, and loud). Subjects were free to choose levels that were most natural for them without any prescribed levels of absolute pitch and loudness (however, subjects were instructed to maintain a constant pitch and loudness within each syllable string). The mean (SD) sound pressure level (10 cm from the lips) of the vowels produced for the soft, comfortable, and loud conditions was 75.5 (4.8), 84.9 (4.9), and 91.8 (4.1) dB SPL, respectively. The mean (SD) of the average fundamental frequency across participants was 250.9 (44.3) Hz.

During syllable production, simultaneous recordings were obtained from the oral airflow using a circumferentially vented high-bandwidth pneumotachograph mask (Glottal Enterprises, Syracuse, NY) with an effective bandwidth of approximately 0 Hz to 1.2 kHz. The airflow signal was low-pass filtered with an 8 kHz 3-dB cutoff frequency (CyberAmp model 380, Axon Instruments, Inc.) and sampled at 20 kHz and 16-bit quantization (Digidata 1440A, Axon Instruments, Inc., Union City, CA). The accelerometer consisted of a miniature piezo-ceramic vibration transducer (BU-27135, Knowles Electronics) with unidirectional sensitivity in the axial dimension and 7.92 mm × 5.59 mm × 4.14 mm footprint. The ACC was wired to a three-conductor cable and potted on a flexible silicone pad with silicone sealant. The sensor was affixed to the anterior neck skin surface halfway between the thyroid prominence and the suprasternal notch along the midsagittal axis using hypoallergenic double-sided tape (model 2181, 3M, Maplewood, MN). As in past studies,[3–5,15] the accelerometer signal was recorded at a sampling rate of 11.025 kHz (16-bit quantization) on a Google/Samsung Nexus S smartphone. Alignment of the oral airflow and accelerometer signals was achieved using a custom MATLAB algorithm that resampled the accelerometer signal to 20 kHz and time-aligned the signal such that the absolute value of the cross-correlation between the accelerometer and acoustic signals was maximized.

As in Holmberg *et al.*,[27] the middle three vowels in each /pae/ string were selected to avoid respiratory initiation and termination effects, yielding a total of nine sets of measurements per loudness condition. The glottal airflow was estimated for one out of the nine vowels based on the vowel that was closest to the mean in terms of acoustic sound pressure level. To avoid onset and offset effects, the mid-50% of each vowel segment was analyzed. Single-formant inverse filtering (a conjugate pair of zeros with unity gain at DC) was applied to a lowpass filtered version (1100 Hz cutoff frequency) of the oral airflow signal for each vowel. The first formant frequency was determined by sweeping candidate frequencies of a single-notch filter with a fixed bandwidth of 70 Hz that minimized formant ripple in the closed phase of the inverse-filtered waveform.[28]

Figure 1 illustrates the estimation of L_1–L_2 in the airflow and neck-surface accelerometer domains as the difference between the amplitudes, in dB, of the first and second harmonics in the frequency spectrum. In the airflow domain, L_1–L_2 was computed from the glottal airflow signal after inverse filtering [Fig. 1(A)]. In the accelerometer domain, L_1–L_2 was computed from the unfiltered, full-bandwidth accelerometer signal [Fig. 1(B)]. Thus, a pair of L_1–L_2 values (from glottal airflow and accelerometer signals) was obtained for each subject per loudness condition.

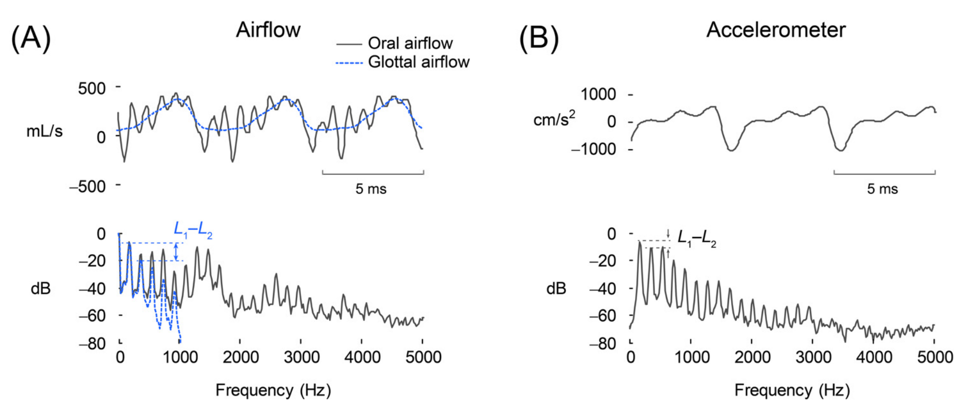

Fig. 1. (Color online) Method of measuring L_1–L_2 from the spectrum of the (A) glottal airflow estimated from oral airflow and (B) anterior neck-surface accelerometer signal.

Two statistical analyses were performed to evaluate whether accelerometer-based L_1–L_2 values correlated with those derived from the glottal airflow signal. First, Pearson's correlation coefficient (r) evaluated the overall relationship between L_1–L_2 measures estimated in the two domains after pooling participant tokens across the three loudness conditions. The alpha level for statistical significance was set at 0.05, and outliers were removed if L_1–L_2 were three standard deviations away from the marginal means. Second, a two-way repeated-measures analysis of variance (rmANOVA) was performed to determine if there were any interaction effects between Signal Type (glottal airflow and accelerometer) and Loudness Condition (soft, comfortable, and loud). A determination of no statistically significant interaction would indicate that the relative relationship of L_1–L_2 values across loudness conditions would be similar in either the glottal airflow or accelerometer domain. In that case, the statistical significance and effect size of the main effects of Signal Type and Loudness Condition would then be quantified using partial eta squared η_p^2. Alternatively, a statistically significant interaction between the two factors would indicate that L_1–L_2 values could not be independently computed in the accelerometer domain.

3. Results

Figure 2 shows a scatterplot of the glottal airflow- and accelerometer-based measures of L_1–L_2 pooling across the three loudness conditions for each speaker. Pearson's correlation coefficient for the relationship was 0.72 (four outliers were identified), with the slope and intercept of the regression line equal to 0.618 dB/dB and 8.6 dB, respectively.

Figure 3 displays box-whiskers plots showing summary statistics of L_1–L_2 within each signal type and loudness condition. These plots are a visual indication of the statistical results of the rmANOVA in Table 1. There was no interaction effect between Signal Type and Loudness Condition ($p = 0.051$). Main effects for both factors were statistically significant, with η_p^2 values equal to 0.373 for Signal Type and 0.334 for Loudness Condition. Thus, although the absolute value of L_1–L_2 was statistically different when computed in the glottal airflow and accelerometer domains (6.9 dB higher in the glottal airflow domain), the relative difference in L_1–L_2 between loudness conditions was maintained regardless of signal type (5.3 dB decrease from soft to comfortable and 2.6 dB decrease from comfortable to loud).

4. Discussion

Results indicated a strong correlation ($r = 0.72$) between L_1–L_2 values computed from the glottal airflow and neck-surface accelerometer signals. The strong correlation between L_1–L_2 in the airflow and accelerometer domains may be due, in part, to this similarity across individuals of the first subglottal resonance (which would primarily affect L_1–L_2), as well as a general lowpass characteristics of normal neck tissue (∼8 dB per octave).[29] The bias between L_1–L_2 values from the accelerometer signal and glottal airflow estimates (6.9 dB higher in the glottal airflow domain) is hypothesized to be largely due to the fact that the left "tail" of the first subglottal resonance amplifies the amplitude of the second harmonic more than the amplitude of the first harmonic.

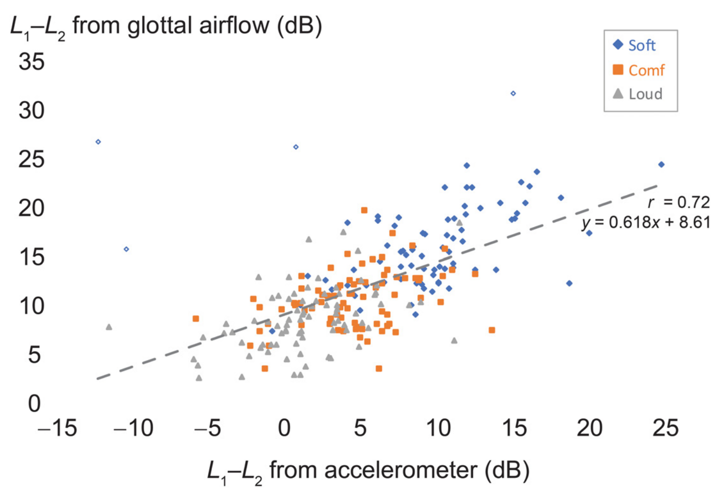

Fig. 2. (Color online) Scatterplot of glottal airflow- and accelerometer-based measures of L_1–L_2 across all participant tokens (pooling across the three loudness conditions of soft, comfortable, and loud).

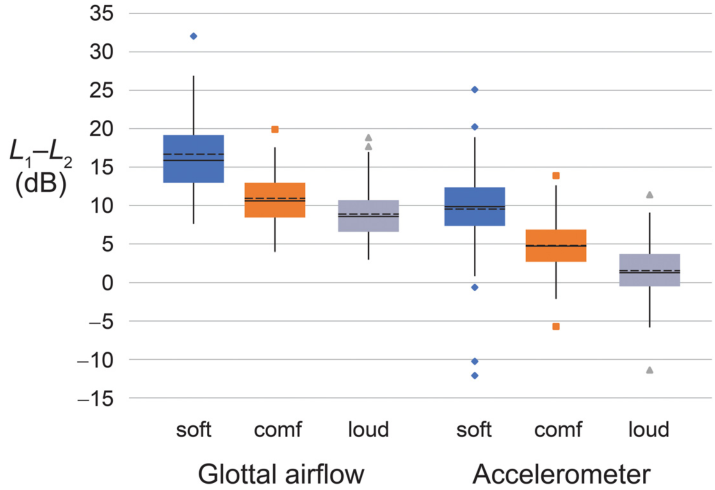

Fig. 3. (Color online) Box-whiskers plots of L_1–L_2 estimated from glottal airflow and anterior neck-surface accelerometer signals within each loudness condition (soft, comfortable, loud).

Future work could elicit various pitches from speakers and sweep the fundamental frequency to test this hypothesis that the subglottal resonances boost L_2 relative to L_1.

At a group level, the first subglottal resonance has been shown to be relatively stable across a sample of adult females with a mean of at 660 Hz and sample standard deviation of 47 Hz.[6] Almost half of the variance between L_1–L_2 values computed from the glottal airflow and neck-surface accelerometer signals remained unexplained (48%), which is hypothesized to be due to across-subject variations in neck morphology, inverse filtering algorithm uncertainty, and, to a lesser extent, subglottal resonances. Thus, comparisons across speakers of accelerometer-based L_1–L_2 are meaningful but should be tempered when assessing voice characteristics. Future work may explore within-speaker relationships between accelerometer waveforms and glottal airflow estimates, which are expected to exhibit lower unexplained variances and be more suitable for detecting changes in underlying vocal function.

The high correlation exhibited by L_1–L_2, in contrast to the weak correlation exhibited by spectral tilt across eight harmonic amplitudes,[15] may be due to less computational sensitivity to higher-frequency spectral effects, such as the number of harmonics within the full spectral bandwidth or mounting of the accelerometer sensor. The current study thus adds L_1–L_2, an interpretable low-bandwidth spectral tilt, to the set of temporal and cepstral measures that have exhibited a high correlation between acoustic/airflow and accelerometer domains. Although correlation does not imply that the same physiological process is being measured, the statistically significant relationships provide evidence that changes in these measures derived from accelerometer signals may be meaningfully interpreted—as they have been in the acoustic and aerodynamic domains—in terms of changes in glottal cycle periodicity, pulse skewness, and open quotient.

Inverse filtering could be performed on the accelerometer signal (to remove effects of the first subglottal resonance) similar to that performed on the oral airflow waveform (to remove effects of the first formant). Alternatively, other methods of inverse filtering, such as impedance-based inverse filtering,[26] could also be employed, but require several steps including aerodynamic calibration and optimization to a vocal system model incorporating subglottal resonances and neck skin properties. These methods were not pursued in this study because of the desire to keep accelerometer-based processing computationally simple. The simpler approach of extracting L_1–L_2 from the raw accelerometer signal is believed to have greater potential for real-time processing that is desirable for, e.g., clinical applications. Ambulatory biofeedback

Table 1. Results of two-way repeated-measures analysis of variance on L_1–L_2 means.

Effect	*df*	η_p^2	*F*	*P*
Signal Type (glottal airflow, accelerometer)	1	0.373	253.9	< 0.001
Loudness Condition (soft, comfortable, loud)	2	0.334	209.7	< 0.001
Signal Type × Loudness Condition	2	0.002	3.0	0.051

technology based on the real-time estimation of accelerometer-based glottal airflow features continues to be developed,[30] with the extraction of frame-based measures of L_1–L_2 from the raw accelerometer waveform computationally efficient using fast Fourier transforms and harmonic peak detection.

5. Conclusion

L_1–L_2 measures extracted from raw neck-surface accelerometer signals and estimates of glottal airflow are highly correlated. Thus, accelerometer-based measures of L_1–L_2 can be generally interpretable as reflecting glottal physiological parameters and voice quality attributes during phonation, making the measures potentially useful for linguistic and clinical voice assessment.

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