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Silent Aspiration Detection by Breath and Swallowing Sound Analysis*

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Abstract— **Detecting aspiration after swallows (the entry of bolus into trachea) is often a difficult task particularly when the patient does not cough; those are called silent aspiration. In this study, the application of acoustical analysis in detecting silent aspiration is investigated. We recorded the swallowing and the breath sounds of 10 individuals with swallowing disorders, who demonstrated silent aspiration during the fiberoptic endoscopic evaluation of swallowing (FEES) assessment. We analyzed the power spectral density (PSD) of the breath sound signals following each swallow; the PSD showed higher magnitude at low frequencies for the breath sounds following an aspiration. Therefore, we divided the frequency range below 300 Hz into 3 sub-bands, over which we calculated the average power as the characteristic features for the classification purpose. Then, the fuzzy k-means unsupervised classification method was deployed to find the two clusters in the data set: the aspirated and non-aspirated groups. The results were evaluated using the FEES assessments provided by the speech language pathologists. The results show 82.3% accuracy in detecting swallows with silent aspiration. Although the proposed method should be verified on a larger dataset, the results are promising for the use of acoustical analysis as a clinical tool to detect silent aspiration.**

I. INTRODUCTION

The act of Swallowing act is one of the most complicated mechanisms in the human body; it involves 25 pairs of muscles in an intricately controlled and coordinated series of events. A normal swallow necessitates an elaborate coordination between the swallowing and the breathing to avoid aspiration, in which the food or liquid enters the airway. Individuals with neuromotor impairments often suffer from swallowing disorders (dysphagia), which includes any swallowing abnormality such as aspiration. There are two methods currently used for the swallowing assessment: the fiberoptic endoscopic evaluation of swallowing (FEES) and the videofluoroscopic swallowing study (VFSS) [1]. However, both techniques are invasive, costly and not convenient for the patients. In recent years, swallowing sound analysis has received considerable attention as a potential non-invasive technique to identify swallowing disorders [2-4]. In this study, we investigated the application of acoustical analysis in detecting silent aspiration.

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Dysphagia can occur in all age groups and as the result of different congenital abnormalities, structural damage, and neurodegenerative diseases [5]. It was observed in 37%-78% of stroke patients and 42%-65% of acquired brain injury (ABI) patients [6]. Aspiration, known as one of the major complications of dysphagia, can result in pneumonia, which contributes to increased morbidity and mortality [7]. The frequency of aspiration after a stroke has been reported to be between 51% and 73% [8], and 38%-41% after ABI [9, 10]. Therefore, when intervening with dysphagic patients, the first goal is to determine whether they are aspirating.

Normally, aspiration is followed by a cough to expel the foreign material from the airway. However, a significant population of dysphagic individuals aspirate silently without demonstrating any clinical sign such as coughing and choking. Thus, these patients' swallowing mechanism should be evaluated by instrumental techniques such as the VFSS or FEES. Both techniques can visualize the structure and function of the swallow by imaging modalities. However, they have some major limitations: the radiation exposure in the VFSS and the invasive nature of the FEES procedure are known as the main disadvantages. Furthermore, the short duration of the study and strict protocols used during these assessments differ greatly from the conditions of a normal meal; that may affect the ability to generalize results to natural setting of patients' normal eating routine. Therefore, an accurate and objective clinical (non-instrumental) assessment of swallowing is essential to detect aspiration, in particular the silent aspirations.

Cervical auscultation was the early attempts towards the swallowing sound analysis [11]; however, this method highly depends on the examiner's skills and thus remained very challenging. In recent years, acoustical analyses of swallowing for detection of dysphagia in general received considerable attention; they have shown a very high sensitivity to distinguish the dysphagic patients from control subjects [3, 12-14]. However, few studies have been reported focusing on the aspiration detection by using the acoustical analysis [15].

In this study, we investigated the potential of the acoustical analysis as a preliminary screening tool that can detect silent aspiration. We hypothesized that the existence of an external particle in the airway, as the result of aspiration must change the sound of flow turbulence during breathing. Hence, we can detect silent aspiration by the sound analysis of the breath sounds immediately after the swallowing event.

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II. METHOD

A. Data

Data were collected from 10 dysphagic adult patients, who suffered from stroke or acquired brain injury. They underwent the FEES assessment as a part of their diagnosis routine. All the patients demonstrated silent aspiration during their FEES procedure. The swallowing sounds were recorded simultaneously with the FEES at Riverview Health Centre, Winnipeg, Canada. The study was approved by the Health Research Ethics Board of the University of Manitoba, and all participants or their legal guardian signed a written consent prior to the experiments.

We recorded the tracheal sounds by a Sony microphone (ECM-88B) placed over the suprasternal notch of the trachea using double-sided tape (Fig. 1). The sounds were recorded at 44.1 kHz sampling rate by a digital sound recorder (EDIROL R-44). Each patient was fed different types of solid and liquid food decided by the speech-language pathologist (SLP) in charge of performing the FEES test. The microphone records both the swallowing and the breath sound signals. Thus, we separated the breath and swallowing signals by aural and visual examination of the signals in the time and frequency domain.

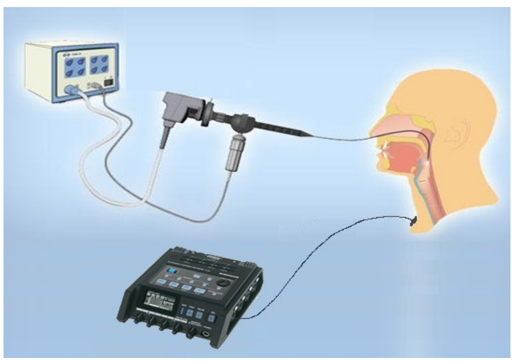

Figure 1. The setup for the swallowing and breath sound recording during the FEES test.

B. Signal Analysis

We selected the features by investigating the signal in both the time and frequency domains. At the first step, we calculated the PSD of the breath signals of patients. Fig. 2 shows the PSDs of the breath sounds following the swallows of a patient; the PSD shows notable variations when there is aspiration; the magnitude of the breath sounds related to aspiration is higher than the others. Also, the variation in the magnitude of the PSD as a result of aspiration is observed to occur at low frequencies (Fig. 2). Therefore, we divided the frequency range below 300 Hz into 3 frequency sub-bands, over which we calculated the average power. Then, for each sub-band we looked into the distribution values.

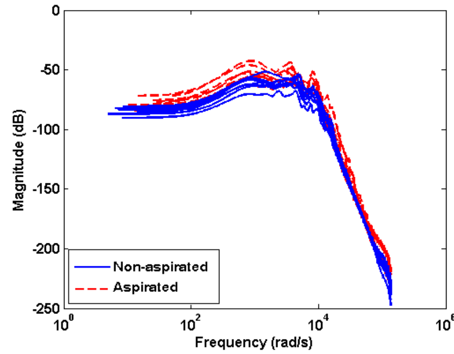

Figure 2. The PSD of breath sound signals after the swallow (simultaneously recorded with FEES) of a dysphagic patients with aspiration (--).

C. Unsupervised Classification

We used the fuzzy k-means clustering algorithm for the unsupervised classification [16]. In this method, each data point belongs to a group of clusters with a membership degree that is optimized through running the algorithm. The membership value is the probability function $\hat{p}(\omega_i|\boldsymbol{X}_j,\hat{\theta})$ whose parameter θ has to be updated by minimizing the cost function as (2):

$$J=\sum_{i=1}^{k}\sum_{j=1}^{n}\left[\hat{p}(\omega_i|\boldsymbol{X}_j,\hat{\theta})\right]^m\left\|\boldsymbol{X}_j-\boldsymbol{V}_i\right\|^2, \qquad (2)$$

where $\boldsymbol{X}_j$ represents the feature vector to be classified, k is the number of clusters, V_i is the vector of cluster centers and m is called the fuzziness index (m>1) [17]. The value of m determines the contribution of the data points with low probability to the cost function. [18].

In this study, we selected $k=2, m=2,$ and $\boldsymbol{X}_j=[P_{av}(60-100), P_{av}(100-200), P_{av}(200-300)]$.

The algorithm begins by initializing the probability value of each data point indicating the degree it belongs to each cluster. The values should meet the condition expressed in (3):

$$\textstyle\sum_{i=1}^{k} p\left(\omega_i|\boldsymbol{X}_j\right)=1. \qquad (3)$$

Then, the center of each cluster is calculated according to (4). Also, the probability of each cluster for every point is updated as:

$$\boldsymbol{V}_i=\frac{\sum_{j=1}^{n} p\left(\omega_i|\boldsymbol{X}_j\right)^m \boldsymbol{X}_j}{\sum_{j=1}^{n} p\left(\omega_i|\boldsymbol{X}_j\right)^m}$$
$$p(\omega_i|\boldsymbol{X}_j)=\frac{\left(1/\left\|\boldsymbol{X}_j-\boldsymbol{V}_i\right\|^2\right)^{1/(m-1)}}{\sum_{i=1}^{k}\left(1/\left\|\boldsymbol{X}_j-\boldsymbol{V}_i\right\|^2\right)^{1/(m-1)}} \qquad (4)$$

The procedures expressed in (4) are repeated until the local minimum of the cost function is achieved. Thus, the centers of clusters are obtained, and every data point is assigned to a cluster where its membership degree has the highest probability.

Having identified the 2 clusters of the breath sounds for each patient, we designed a heuristic screening scheme to classify those breaths after an aspirated swallow as the following: If any of the breaths (up to 3) after a swallow belongs to the aspirated cluster, then that swallow is labeled as an aspirated swallow. To calculate the accuracy, specificity and sensitivity, true positives were defined as the swallows marked as aspiration, and true negatives are the non-aspiration swallows.

III. Results

We applied the fuzzy k-means clustering algorithms to the 3 sub-bands average power feature vectors. The breath sounds after the swallows resulted in 2 clusters as shown in Fig. 3; this pattern was observed for all patients. Comparing with the patient's FEES evaluation results, we found that the majority of the data points of the cluster formed near the origin belonged to swallows without aspiration, while the other cluster had the majority of the aspirated swallows. We summarized the results of the unsupervised clustering of the swallows of all the patients in Table I.

Some patients showed a few number of aspirated swallows. Therefore, the average accuracy may not be an appropriate representative of the method's accuracy. Thus, we have shown the classification results in Table I in terms of the number of True Positive (TP), False Positive (FP), True Negative (TN), and False Negative (FN). The average accuracy, specificity and sensitivity were found to be 82.3%, 81.4%, and 84.8%, respectively. For 5 out of 10 patients the sensitivity of the method was found to be 100% as all the aspirations were detected correctly.

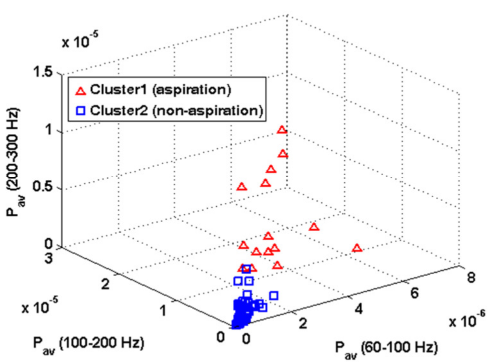

Figure 3. The unsupervised classification results of the three average power features of the breath sounds after swallows of a patient.

TABLE I. The classification results for the aspiration detection.

No. of swallows	No. of swallows with aspiration	TP	FP	TN	FN
186	46	39	26	114	7

IV. Discussion

The idea of investigating the frequency features of the breath sounds originally comes from the fluid mechanics characteristics of airway dynamics through which the airflow (expiration/inspiration) occurs. It has been shown that the airflow in the trachea is highly likely to be turbulent [19]. Thus, any particle in the airway will increase the size of the turbulence large scales, which results in a higher power at low frequencies [20]. Aspiration may change the pattern of the flow by creating vortices. Assuming the particle as a circular cylinder, we can calculate the frequency of vortices formed in the airway, f, by the von Karman formula [21]:

$$\frac{fd}{U} = 0.198\left(1 - \frac{19.7}{Re}\right), \tag{5}$$

where U is the instantaneous speed of airflow, d is the diameter of the particle, and Re is the Reynolds number. It was shown that the typical inspiratory speed varies between 0.68 and 2.70 m/s [22] and Re varies between 800 in light breathing and 9300 in heavy breathing. Therefore, for a small particle with an average diameter of 1 mm the frequency of vortices can be calculated to be in the range of 136-540 Hz. This is congruent with the frequency range that the features are selected in this study.

Physicians have used cervical auscultation to detect aspiration; however, there has not been any other study that analyzed the tracheal sounds to detect aspiration. Our proposed method outperforms the cervical auscultation studies for detecting aspiration [23, 24]. The results show 84.8% sensitivity and 81.4% specificity for silent aspiration detection compared to sensitivity and specificity values of 62% and 66% [24] and 94% and 70% [23], who used cervical auscultation for aspiration detection. Our method has overcome the overestimation issue (high false positive error) which was the case in those studies.

Despite the small sample size, the results of this study are encouraging. Once these results are confirmed on a larger dataset, the proposed method may lead to development of a non-invasive and reliable supplemental diagnostic/screening tool to the clinical examination of swallowing function that can be performed at the patient's bedside to detect silent aspiration and determine the appropriate dietetic treatment for the patient.

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