

# DECISION-TREE BASED ANALYSIS OF SPEAKING MODE DISCREPANCIES IN EMG-BASED SPEECH RECOGNITION

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**Abstract:** This study is concerned with the impact of speaking mode variabilities on speech recognition by surface electromyography (EMG). In EMG-based speech recognition, we capture the electric potentials of the human articulatory muscles by surface electrodes, so that the resulting signal can be used for speech processing. This enables the user to communicate silently, without uttering any sound. Previous studies have shown that the processing of silent speech creates a new challenge, namely that EMG signals of audible and silent speech are quite distinct. In this study we consider EMG signals of three speaking modes: audibly spoken speech, whispered speech, and silently mouthed speech. We present an approach to quantify the differences between these speaking modes by means of *phonetic decision trees* and show that this measure correlates highly with differences in the performance of a recognizer on the different speaking modes. We furthermore reinvestigate the *spectral mapping* algorithm, which reduces the discrepancy between different speaking modes, and give an evaluation of its effectiveness.

## 1 INTRODUCTION

The past decades have seen rapid advancements in all areas of automatic speech processing, including speech recognition, spoken language translation, and speech synthesis. With these developments, the use of speech and language technologies has become increasingly common in a large variety of applications, such as information retrieval systems, voice-operated cell phones, call center services, car navigation systems, personal dictation and translation assistance, as well as applications in military and security domains.

Despite these achievements, speech-based interfaces working with acoustic speech signals still have several limitations, i. e. the performance degrades significantly when environmental noise is present, and communication in public places is difficult since bystanders may be disturbed and confidentiality is breached by the clearly audible speech. Also, speech-disabled people may be unable to use voice-controlled systems.

A relatively novel approach to address these challenges is the application of Silent Speech Interfaces (SSI), which are electronic systems enabling communication by speech without the necessity of emitting an audible acoustic signal (Denby et al., 2010). In this paper, we report on our most recent results in electromyographic (EMG) speech recognition, where the

activation potentials of the articulatory muscles are directly recorded from the subject's face via surface electrodes.

Automatic recognition of silent speech by means of electromyography is currently heavily investigated, and the performance becomes good enough to allow for communication applications (Wand and Schultz, 2011). The research focus of this study is the variation of the EMG signal for different *speaking modes*, in particular, for audibly spoken versus silently mouthed speech. A first series of experiments was reported (Wand et al., 2009; Janke et al., 2010a; Janke et al., 2010b), where the authors investigated the differences between audibly and silently spoken speech based on the *Power Spectral Density* (PSD) of raw EMG signals. This quantity is an estimate of the energy an EMG signal contains. Those experiments showed that the PSD of EMG signals is lower for silent speech than for audibly spoken speech, however, for phones which create a relatively high *sensorimotor feedback* when articulated, the PSD of this EMG signal segment will also be relatively high. It could also be shown that for talented silent speakers with similar recognition accuracies across all speaking modes, the signal energy discrepancy is much smaller than for less talented silent speakers.

In (Wand et al., 2011), the authors present a different approach for studying the discrepancy be-

tween audible and silent speech, namely by means of *phonetic decision trees*. Phonetic decision trees are commonly used in almost all state-of-the-art speech recognition systems to model the effects of coarticulation. We use phonetic decision trees as part of our *Bundled Phonetic Feature* modeling (see section 3). In (Wand et al., 2011), the results of the tree splitting process are used as a diagnostic tool to explore the impact of speaking mode dependencies on the phonetic models of the EMG-based speech recognizer.

This study leverages off and improves this method. In our first experiment, we investigate the *entropy gains* which are associated with the decision tree splitting process. We show that they give a measure for the discrepancy between audible and silent EMG, and that this measure remains stable even when *spectral mapping* (Janke et al., 2010a) is applied. We compare the results to previously developed speaking mode discrepancy measures (Wand et al., 2009; Janke et al., 2010a) and show that they correspond well with each other.

As a second experiment, we perform a detailed investigation on the EMG signals of *whispered* speech. We show that whispered speech accords well with audible speech, but also show that the spectral mapping algorithm gives further improvements.

The remainder of this paper is organized as follows: Section 2 describes our corpus, and section 3 outlines the structure of the EMG-based speech recognizer and describes the phonetic feature bundling algorithm. Sections 4 reviews the state-of-the-art methods to describe speaking mode variabilities by means of the EMG signal, and explains our new method. In sections 5 and 6, we apply our method to EMG signals of silent speech and whispered speech, respectively. Finally, section 7 concludes the paper.

## 2 DATA CORPUS

For this study we used a subset of the EMG-UKA corpus (Janke et al., 2010a), namely the subset of recording sessions which contain EMG signals of audible, whispered, and silent speech. The following description is based on (Wand et al., 2011).

The corpus of this study consists of eight speakers, each of whom recorded between one and eleven sessions, resulting in a total amount of 25 sessions<sup>1</sup>. The recording protocol was as follows: In a quiet room, the speaker read 50 English sentences for three times, first audibly, then in whispered speech, and at

<sup>1</sup>Note that we added one session which was not present in the corpus for (Wand et al., 2011)

last silently mouthed. As an abbreviation, we call the EMG signals from these speaking modes *audible EMG*, *whispered EMG*, and *silent EMG*, respectively.

Each part consists of a BASE set of 10 sentences which were identical for all speakers and all sessions, and one SPEC set of 40 sentences, which varied across sessions. However in each session, these sentence sets were the same for all three parts, so that the database covers all three speaking modes with parallel utterances. The total of 50 BASE and SPEC utterances in each part were recorded in random order. In all recognition experiments, the 40 SPEC utterances are used for training, and the 10 BASE utterances are used as test set.

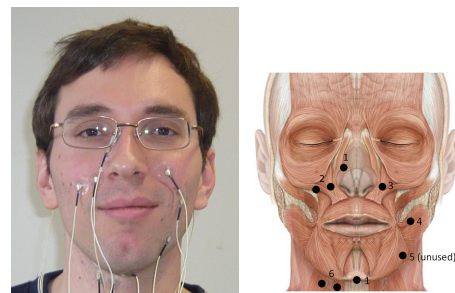


Figure 1: Electrode positioning (Wand et al., 2011) (muscle chart adapted from (Schünke et al., 2006)).

The EMG recording was done with a computer-controlled 6-channel EMG data acquisition system (Varioport, Becker-Meditec, Germany). All EMG signals were sampled at 600 Hz. The electrode setting uses six channels and captures signals from the levator angulis oris (channels 2 and 3), the zygomaticus major (channels 2 and 3), the platysma (channel 4), the anterior belly of the digastric (channel 1) and the tongue (channels 1 and 6). Channels 2 and 6 use bipolar derivation, whereas channels 3, 4, and 5 were derived unipolarly, with two reference electrodes placed on the mastoid portion of the temporal bone (see Figure 1). Similarly, channel 1 uses unipolar derivation with the reference electrode attached to the nose. Channel 5 remains unused since the signal is unstable and contains too many recording artifacts.

In the audible and whispered parts, the audio signal was simultaneously recorded with a standard close-talking microphone connected to a USB sound-card. The audio sampling rate was 16 kHz.

The total corpus for this study consists of the following amounts of data:

	Data length (minutes)		
	Training	Test	Total
Audible EMG	69	19	88
Whispered EMG	71	20	91
Silent EMG	70	19	89
All modes	210	58	268

### 3 RECOGNIZER SETUP

#### 3.1 Feature Extraction

The feature extraction is based on *time-domain features* (Jou et al., 2006). Here, for any given feature  $\mathbf{f}$ ,  $\bar{\mathbf{f}}$  is its frame-based time-domain mean,  $\mathbf{P}_{\mathbf{f}}$  is its frame-based power, and  $\mathbf{z}_{\mathbf{f}}$  is its frame-based zero-crossing rate.  $S(\mathbf{f}, n)$  is the stacking of adjacent frames of feature  $\mathbf{f}$  in the size of  $2n + 1$  ( $-n$  to  $n$ ) frames.

For an EMG signal with normalized mean  $x[n]$ , the nine-point double-averaged signal  $w[n]$  is defined as

$$w[n] = \frac{1}{9} \sum_{k=-4}^4 v[n+k], \quad \text{where} \quad v[n] = \frac{1}{9} \sum_{k=-4}^4 x[n+k].$$

The high-frequency signal is  $p[n] = x[n] - w[n]$ , and the rectified high-frequency signal is  $r[n] = |p[n]|$ . The final feature **TD15** is defined as follows:

$$\mathbf{TD15} = S(\mathbf{f2}, 15), \text{ where } \mathbf{f2} = [\bar{\mathbf{w}}, \mathbf{P}_{\mathbf{w}}, \mathbf{P}_{\mathbf{r}}, \mathbf{z}_{\mathbf{p}}, \bar{\mathbf{r}}].$$

As in (Jou et al., 2006; Wand et al., 2009), frame size and frame shift were set to 27 ms respective 10 ms. In all cases, we apply LDA on the **TD15** feature to reduce it to 32 dimensions. The feature extraction may be preceded by application of the spectral mapping algorithm, see section 4.1 for a description of this algorithm.

#### 3.2 Bootstrapping the EMG-based Speech Recognizer

For training the recognizer, we need precise phonetic *time-alignments* of the training data set.

For *audible* and *whispered* EMG, these alignments can be relatively easily obtained by processing the parallelly recorded acoustic data with a conventional speech recognizer. The setup of this recognizer is described in (Jou et al., 2006). For *silent* EMG, this method is impossible since no acoustic data exists. Our method to obtain initial time-alignments for the silent EMG data works by first training a recognizer with the audible EMG training data of the same session, and then applying this recognizer to create time-alignments for the silent EMG data. This method is called *Cross-Modal Labeling* (Wand et al., 2009).

#### 3.3 Phonetic Feature Bundling

The EMG-based speech recognizer is based on three-state left-to-right fully continuous Hidden Markov Models (HMMs). In contrast to most state-of-the-art *acoustic* speech recognition systems, the HMM states

are modeled not with phonemes, but with binary-valued *phonetic features* (PFs) (Kirchhoff, 1999), which represent articulatory properties of a given phoneme, such as the place or the manner of articulation. Note that in earlier works, phonetic features are also called *articulatory features*.

The architecture of the PF-based EMG decoding system is a *multi-stream* architecture (Metze and Waibel, 2002): the feature vector models draw their emission probabilities not from one single source, but from a weighted sum of various sources which correspond to Gaussian mixture models representing sub-states of PFs.

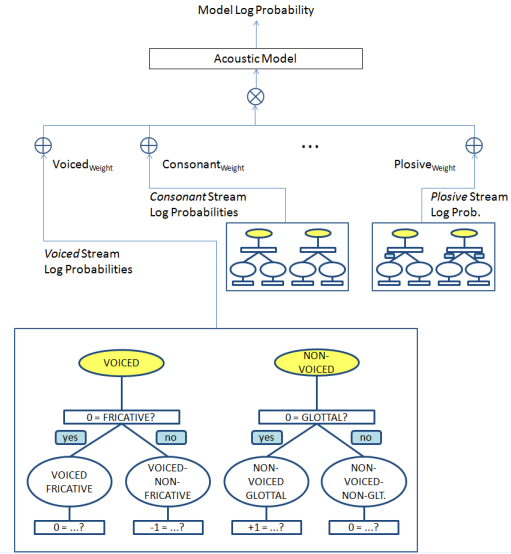


Figure 2: Structure of a multi-stream recognizer based on Bundled Phonetic Features (BDPFs). Our system uses the nine most common phonetic features as root nodes of the BDPF trees. Note that substates (begin, middle, end) are not shown.

*Phonetic feature bundling* (Schultz and Wand, 2010) is the process of pooling dependent features together, so that in the end one obtains a set of models which represent *bundles* of PFs, like “voiced fricative” or “rounded front vowel”. These models are called *Bundled Phonetic Features* (BDPF), they play a central role in our recognition system.

This pooling is performed by phonetic decision trees, which implement a data-driven approach to incrementally create specific models from general ones (Bahl et al., 1991). The basic idea is to go from general models to specific models by *splitting* a model based on a phonetic question. Phonetic feature bundling starts with seven models for *each stream*. The models represent the beginning, middle, and end states of the present or absent phonetic feature, plus a silence state. The splitting questions cover both the current phone and the left and right context phones.

The criterion for the choice of the splitting question in each step is the information gain or entropy loss (Finke and Rogina, 1997). The splitting process stops at a fixed number of 80 leaves for each tree, which on average is the optimal number of leaves for this corpus. Our system uses nine streams, which are based on the nine most frequent phonetic features in the EMG-UKA corpus: Voiced, Consonant, Vowel, Alveolar, Unround, Fricative, Unvoiced, Front, Plosive.

Figure 2 shows a graphical overview of the model structure of this recognizer. (Schultz and Wand, 2010) reported that a recognizer based on bundled phonetic features outperforms a context-independent phone-based system by more than 30% relative. On the EMG-UKA corpus, the best average word error rate of this recognizer when trained on *audible* utterances is 24.12% (Wand et al., 2011).

The system allows to augment phonemes with additional attributes, such as speaker properties or the speaking mode. Our investigation of silent and whispered speech based on decision tree properties leverages off this method, see section 4.2 for the experiment description and the results.

### 3.4 Training and Decoding

We trained EMG-based speech recognizers for each recording session of each speaker (*session-dependent* recognizers). The recognizers were trained on EMG data from two speaking modes, where the training data consisted of 40 SPEC sentences per speaking mode (see Section 2).

For decoding we used the trained acoustic model together with a trigram Broadcast News language model giving a perplexity on the test set of 24.24. The decoding vocabulary was restricted to the 108 words appearing in the test set, which in previous works (see i. e. (Schultz and Wand, 2010)) is the standard procedure for small session-dependent systems. We applied lattice rescoring to obtain the best weighting of language model and acoustic model parameters.

Figure 3 gives a breakdown of the recognition results of a *mode-independent* EMG-based speech recognizer trained on both audible and silent EMG data. We trained session-dependent recognizers on all 25 sessions of the corpus and then averaged over all sessions of each speaker. The average WER over all 25 sessions on audible EMG is 26.39%, while the average WER on silent EMG is 48.32%.

From Figure 3 we observe that (1) the performance difference between silent and audible mode is significant and (2) the performance varies greatly across speakers. We also see that most speakers perform consistently across sessions.

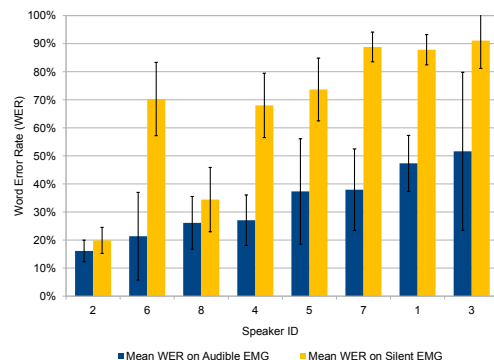


Figure 3: Baseline Word Error Rates for audible and silent EMG on a mode-independent recognizer. The error bars give the confidence interval at a 5% confidence level.

## 4 DESCRIBING THE VARIABILITY OF SPEAKING MODES

In this section, we describe in detail how we extend the BDPF algorithm introduced in section 3.3 to gain insights into the variability of speaking modes. We start with a review of previously developed methods to quantify the discrepancy between audible, silent, and whispered speech, and then lay out how our decision-tree based method augments and improves those results.

As a baseline measure for the discrepancy between the EMG data of different speaking modes in a particular session, we use the *difference* between the Word Error Rates of an EMG-based recognizer on these speaking modes. The computation e. g. for audible and silent speech works as follows: For a given session, we train a *mode-independent* recognizer on the training data sets of audible and silent EMG, and then separately test the recognizer on the audible and silent EMG test sets. The difference between the two Word Error Rates is then used as our measure.

The rest of this section proceeds as follows: We firstly give a review of the Power Spectral Density Method for evaluating the discrepancy between different speaking modes (Janke et al., 2010a; Janke et al., 2010b), and outline the spectral mapping algorithm which has been crafted to reduce this discrepancy. We then describe our new method to evaluate the speaking mode discrepancy. The experiments and results are given in sections 5 and 6.

#### 4.1 Review of Power Spectral Density Methods

In a preliminary experiment (Wand et al., 2009), it was shown that typically, EMG signals of silent speech have lower power than EMG signals of audible speech. A more versatile investigation was performed in (Janke et al., 2010a; Janke et al., 2010b), where the *Power Spectral Density (PSD)* was used as a measure for the variation between EMG signals of different speaking modes. The Power Spectral Density is a measure of the energy which the EMG signal contains at different frequency ranges. The computation follows Welch's method (Welch, 1967), which proceeds as follows (Janke et al., 2010a):

- The input signal is divided into windows with a length of 30 samples, with 67% overlap.
- Each segment is windowed with a Hamming window to reduce spectral distortion.
- On each segment, a 256-point Fast Fourier Transform is computed, yielding a Fourier representation of the signal.
- The square magnitude of the FFT is taken and averaged over all segments, yielding the Power Spectral Density (PSD) per utterance.

The PSDs are then averaged over all training utterances of a particular session and speaking mode. The result of this computation is a set of three PSD curves representing the spectral contents of the audible, whispered, and silent EMG part of the given session.

Figure 4 shows two exemplary PSD curves of the EMG channel 6 for audible, whispered, and silent speech. The above part shows the PSDs for the first session of Speaker 1, who has moderate skills in speaking silently. The curve shapes look similar, but the amplitudes differ for the speaking modes: In particular, the PSD of silent EMG is always much lower than the PSD of audible EMG. The PSD curve for whispered speech is situated in-between the curves for audible and silent EMG. Evaluated on separate recognizers for audible, whispered, and silent EMG, respectively, this speaker has a Word Error Rate (WER) of 57.6% on audible EMG, while on silent EMG the WER is 92.9%. Whispered speech is recognized with 62.6% WER.

The lower part charts the PSD curves of a well practiced silent speaker (Speaker 2) with good recognition rates for all speaking modes. The shape of the PSD curves is somewhat different from those of speaker 1, but in particular, the curves are much closer together. Consequently, the WERs for the

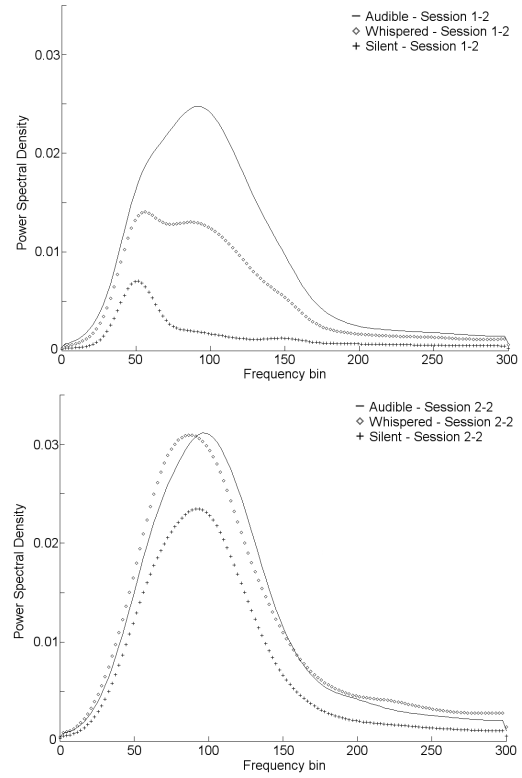


Figure 4: PSD of EMG channel 6 of a novice silent speaker (above) and an experienced silent speaker (below) for audible, whispered and silent speech.

three speaking modes are much more similar: Audible EMG is recognized with 17.2% WER, whispered speech is recognized with 19.2% WER, and silent speech is recognized with 18.2% WER.

This observation indicates that there is some relationship between the ratio of the PSD curves for different speaking modes and the WER difference between different speaking modes. In order to quantify this statement, one can use the *PSD Ratio* between speaking modes, which is determined by computing the ratio of audible EMG and silent EMG PSD for each channel and each frequency bin and taking the maximum of this ratio over the frequency bins. (Janke et al., 2010a) reports a correlation between PSD ratio and WER difference of 0.72. In this study, we find correlations of about 0.5, depending on the EMG channel the PSD ratio is computed on. This disparity may be due to our slightly different recognizer setup: (Janke et al., 2010a) train independent recognizers for audible and silent EMG, whereas we train mode-independent recognizers on training data from both audible and silent EMG.

Based on the PSD ratio measure, (Janke et al., 2010a) develops the *spectral mapping* algorithm which reduces the discrepancy between audible and

silent EMG. The spectral mapping algorithm is applied to raw EMG signals of *silent* speech and works as follows:

1. One computes the channel-wise Power Spectral Density (PSD) ratio of the silent EMG signals and audible EMG signals, as a function of the frequency. The result is averaged over all utterances of one session. This frequency-dependent ratio is called the *mapping factor*.
2. Each silent EMG utterance is transformed into the frequency domain by the Fast Fourier Transform (FFT), then every frequency component is multiplied with the corresponding mapping factor, and the resulting transformed frequency representation of the signal is transformed back into the time domain by the inverse FFT.
3. After this procedure, features are extracted from the transformed signal as usual.

In (Janke et al., 2010a), it is shown that the spectral mapping algorithm reduces the WER of a silent EMG recognizer trained by the Cross-Modal Labeling approach by about 7% relative. When a recognizer is trained on audible EMG and tested on silent EMG without an intermediate training on Silent EMG data (*Cross-Modal Training*), the improvement is much higher, at about 12% relative.

## 4.2 Evaluating the Speaking Mode Discrepancy based on a Polyphone Decision Tree

In this paper we propose a new method to evaluate speaking mode discrepancies between different speaking modes. This method is based on our BDPF framework (see section 3.3) and draws its validity from the fact that BDPF bundling splits Gaussian mixture models in a *data-driven* manner without resorting to any kind of prior knowledge or assumption. Our approach is to tag each phone of the training data set with its speaking mode (audible or silent). We then let the decision tree splitting algorithm ask questions about these attributes.

While we do use the resulting models for our EMG-based speech recognizer, our main target is a different one: We follow the approach from (Schultz and Waibel, 2001) and examine the *entropy gains* associated with the model splitting process.

The details are as follows: For each phonetic feature, the tree splitting process starts out with six root nodes (plus a silence node, which is not considered for splitting). In each step, one question is asked, and one existing node is split, creating two new nodes.

This process is repeated until a stopping criterion is met.

Each node split is associated to one question and one entropy gain, which is the maximum entropy gain possible at this step. When this algorithm has terminated, there appear two kinds of questions:

- Questions about phonetic features (see section 3.3)
- Questions about the speaking mode.

Clearly, questions about the speaking mode may appear in several places across the tree, but the algorithm forbids a question which has already been asked to be repeated in any descendant node.

We now calculate the total entropy gain when traversing the model tree. Figure 5 plots the entropy gain for speaking mode questions and phonetic feature questions over the total number of questions asked, for a speaker where the discrepancy between audible and silent speech is relatively large (above) respectively relatively small (below). The values are averaged over all nine PF trees. It can be seen that in the latter case, the speaking mode questions do not contribute much to the entropy gain at all, while in the first case, the speaking mode questions are responsible for a large amount of the entropy gain. However when more and more questions are asked, there is hardly any further gain associated to speaking mode questions. Note that we never normalize the entropy gain.

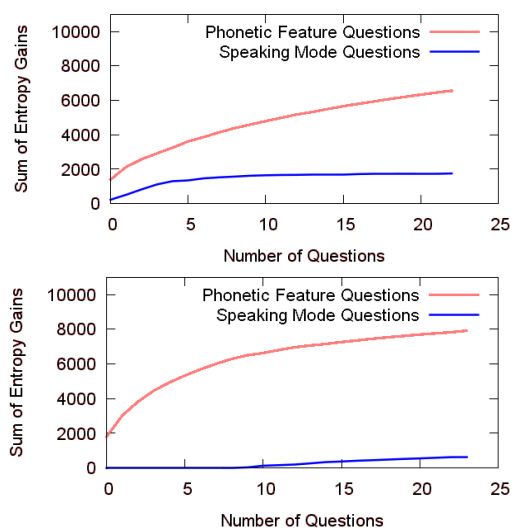


Figure 5: Entropy gains for a speaker with high discrepancy (top)/low discrepancy (bottom) between audible and silent EMG, plotted over the number of splitting questions asked.

Based on this observation, we now consider all PF trees and look at the one question which yields the *highest* entropy gain of all questions about the

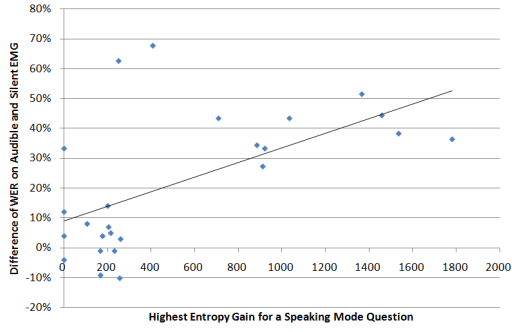


Figure 6: Scatter plot of the Maximum Entropy Gain and the WER difference between silent and audible EMG, with regression line.

speaking mode. We use this *maximum entropy gain* as a measure for the discrepancy between speaking modes: If there is hardly any difference between speaking modes, the maximum entropy gain should be small, possibly even zero if no speaking mode question at all has been asked. If the EMG signals of different speaking modes differ a lot, there should be a high entropy gain associated to a speaking mode question.

In (Wand et al., 2011), the authors present another measure for the discrepancy between speaking modes based on phonetic decision trees: In the final decision tree, the number of tree leaves dependent on the speaking mode is counted. The fraction of “mode-dependent tree nodes” (MDN) out of the set of all nodes is then used as a measure for the speaking mode discrepancy.

In sections 5 and 6, we will compare our entropy-based method to the MDN method and show that they yield similar results. One advantage of the entropy gain measure is a *local* property of the decision trees: When the decision tree gets larger and larger, the maximum entropy gain does not change any more. In contrast to this, the fraction of mode-dependent nodes may still change when the decision tree gets larger, so in order to get a stable measure here, one must fix a stopping criterion for the tree splitting.

## 5 EVALUATION ON SILENT EMG

As a first experiment, we take the baseline system from section 3. We train session-dependent systems on all sessions of our corpus, using both the audible and the silent EMG training data set. The average WER over all sessions is 26.39% for *audible* EMG and 48.32% for *silent* EMG.

Figure 5 depicts typical entropy gain curves for two different speakers: In the upper plot, we have a

speaker with a high discrepancy between audible and silent EMG, in the lower plot, there is a speaker with a low discrepancy. One can see that the entropy gains are much larger in the first case than in the second case, and that after a relatively small number of questions, the entropy gain does not change a lot any more since the differences between modes are clustered out.

This observation suggests to use the maximum entropy gain as a discrepancy measure between audible and silent EMG, as described in section 4.2. The maximum entropy gain varies across sessions from zero to 1781, with an average of 530, and correlates with the WER difference between audible and silent EMG with a correlation coefficient of 0.59. This is a higher correlation than for the PSD ratio, where we only obtain a correlation coefficient of 0.53.

We can also compare the maximum entropy gain and the percentage of mode-dependent nodes, which has been described in (Wand et al., 2011). Over all sessions of our corpus, the percentage of mode-dependent nodes varies between 0% and 95%, with an average of 36%, it also correlates with the WER difference with a correlation coefficient of 0.59.

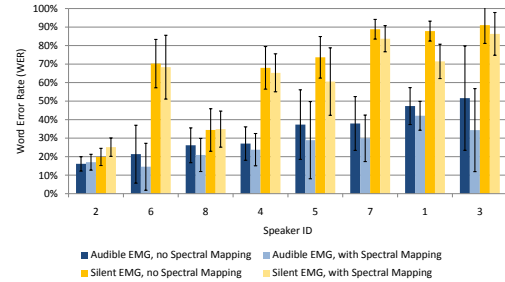


Figure 7: Word Error Rates of an EMG-based Speech Recognizer trained on audible and silent EMG, with and without spectral mapping. The error bars give the confidence interval at a 5% confidence level.

Figure 6 shows a scatter plot of the maximum entropy gain versus the WER difference. It can be seen that the sessions fall into two main categories: On the one hand, sessions with a very low WER difference and, consequently, a very low maximum entropy gain. On the other hand, sessions with a higher WER difference and a high entropy gain.

As a second experiment, we applied spectral mapping (see section 4.1) to the silent EMG data before training and testing the recognizer. As expected, the maximum entropy gain drops from an average of 530 to an average of 294. The correlation between maximum entropy gain and WER difference remains relatively high, the correlation coefficient is 0.43.

Figure 7 gives a full breakdown of the average recognition results by speaker for audible and silent EMG, with and without spectral mapping. One can

observe that Spectral Mapping almost always yields an improvement in both speaking modes, with the sole exception of the very best speaker.

## 6 EVALUATION ON WHISPERED EMG

Up to now, most research work has focused on the discrepancy between the audible and silent EMG speaking modes. Whispered EMG is considered in some works (Janke et al., 2010a; Janke et al., 2010b), but is not the main focus.

In this section, we train a recognizer on data from both the audible and the whispered speaking mode. We show that whispered EMG and audible EMG are well compatible with each other, but that despite that, spectral mapping applied to whispered EMG still yields some improvement. We particularly investigate the entropy gain associated with splitting models according to questions about the audible and whispered speaking mode.

As a first experiment, we take the trained Gaussian mixture models and compute the entropy gains associated with questions about the speaking mode. It turns out that in our corpus, for most of the speakers there is little difference between the audible and the whispered speaking mode, which can be deduced from the fact that there is hardly any entropy gain associated with questions about the speaking mode.

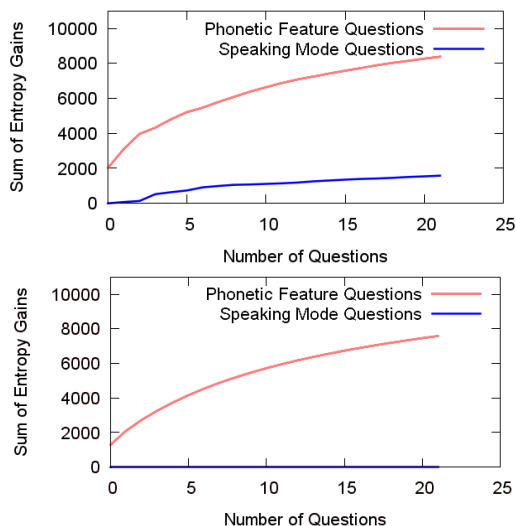


Figure 8: Entropy gains for a speaker with high discrepancy (above)/low discrepancy (below) between audible and whispered EMG, plotted over the number of splitting questions asked.

Figure 8 shows the entropy gains for phonetic fea-

ture questions and speaking mode questions plotted over the total number of questions asked. For speakers 2 to 8, the curves generally look similar to the lower one—asking speaking mode questions yields practically no entropy gain. However for speaker one, the discrepancy between audible and whispered EMG is quite high (see the above figure). Listening to the audio recordings of speaker 1 shows that this speaker has indeed a very quiet, almost inaudible way of whispering, which may explain the observed discrepancy.

The mode-independent recognizer for audible and whispered EMG attains a recognition rate of 23.19% for audible EMG and 21.79% for whispered EMG, averaged over all 25 sessions. It is instructive to compare these recognition rates to a system which has been trained exclusively on the audible or whispered EMG training data set: In this case, audible EMG is recognized with 29.74% WER, and whispered EMG is recognized with 30.80% WER, on average. While these two numbers are just as close as for the mode-independent recognizer, we see that combining audible and whispered EMG data, which doubles the size of the training data set, brings a large WER improvement of more than 23% relative, which is just what one expects when increasing the size of the training data set with *consistent* data.

Finally, we apply spectral mapping to the whispered EMG data. Just like in the case of silent EMG data, we compute frequency-dependent mapping factors between whispered EMG and audible EMG for each speaker. We then take each whispered EMG utterance, transform it into the Fourier domain, multiply each frequency component with the corresponding mapping factor, and back-transform the modified signal into the time domain.

This process improves the average WER on *audible* EMG by about 6% relative, from 23.19% to 21.74%. The average WER on whispered EMG remains unchanged. The result gets clearer if one considers a breakdown of the improvement by speaker: It turns out that indeed, for speaker 1 with a large discrepancy between audible and whispered EMG, there is a significant and large improvement for both speaking modes. For the other speakers, the results show more variation; in some cases the recognition result gets worse when spectral mapping is applied.

## 7 CONCLUSIONS

In this paper, we proposed a new method to evaluate the discrepancies between the audible, whispered, and silent speaking modes in EMG-based speech recognition. We considered the phonetic decision trees which

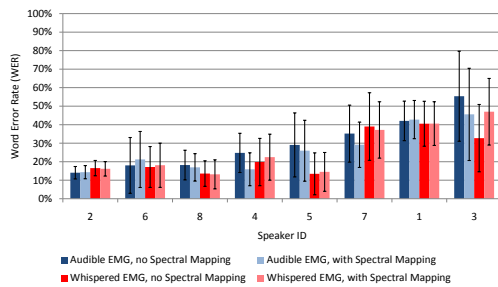


Figure 9: Word Error Rates of an EMG-based Speech Recognizer trained on audible and whispered EMG, with and without spectral mapping. The error bars give the confidence interval at a 5% confidence level.

are created during the training process of the recognizers, allowed the decision tree creation algorithm to split tree nodes according to the speaking mode, and then considered the *entropy gains* which are associated with tree node splits due to a speaking mode question.

We showed that the differences in silent speaking style between speakers may be drastic, and that an evaluation of the decision tree entropy gains well characterizes the speaker's ability to speak silently. Comparing audible speech to silently mouthed speech, we ascertained that the maximal entropy gain which is due to a speaking mode question may be used as a measure for the discrepancy between speaking modes, and that this measure remains stable even when the spectral mapping algorithm is applied.

Building upon this, we trained, for the first time, an EMG-based speech recognizer on EMG recordings of both audible and *whispered* speech. It turned out that whispered speech is, for most speakers, quite compatible to audible speech, but that in the EMG-UKA corpus, there is one speaker where the discrepancy between audible and whispered speech is quite large. We also showed that some accuracy gain can be achieved with the spectral mapping algorithm.

Based on our decision tree analysis method, possible future work includes a more detailed phonetic analysis of the discrepancy between audible and silent speech, as well as the improvement of the spectral mapping algorithm to take phone information into account.

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